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The effect of localized muscle fatigue on tibial impact acceleration.

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The effect of localized muscle fatigue on tibial impact acceleration

by

Janice Flynn

A Thesis

**Submitted to the Faculty of Graduate Studies and Research
Through Human Kinetics
in Partial Fulfillment of the Requirements for
the Degree of Master of Human Kinetics at the
University of Windsor**

Windsor, Ontario, Canada

2003

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ABSTRACT

Tibial impact acceleration measured during running is known to increase with general body fatigue. The purpose of this study was to determine if localized muscle fatigue of the shank muscles would also cause an increase in peak tibial acceleration. The human pendulum system was used to control impact velocity and joint angle. There were 24 women participating in the study: 12 between the ages of 20 – 25, and 12 between 50 – 60 years. Each lay supine on the pendulum, and their unshod dominant heel was impacted into a vertical force plate with a velocity between 1 – 1.15 m/s, and a force approximating 1.8 – 2.8 X body weight. A uni-axial accelerometer at the tibial tubercle measured peak tibial acceleration, time to peak acceleration and the slope of the acceleration/time curve. EMG activity of the tibialis anterior and gastrocnemius was used to define and monitor fatigue. The dorsiflexors and plantarflexors were fatigued on two separate days, at least a week apart. Statistical analysis revealed a significant decrease in peak acceleration and acceleration slope following fatigue. There were no significant main effects or interactions for age group or muscle group. In conclusion, localized muscle fatigue of the dorsiflexors or plantarflexors resulted in a significant decrease in the peak acceleration and acceleration slope measured at the knee, which is opposite to the effect of general body fatigue.

DEDICATION

This thesis is dedicated to three people who have had the greatest influence over my choices in life. Firstly, to my parents, Marge and Ivan Moreside. They taught me, probably unknowingly, to seek out and embrace a challenge. Secondly, to my husband, Greg, who has provided me with constant support and encouragement for 27 years. Thanks to him, I realize that mid-life is not necessarily a time to slow down and reflect on the past, but rather a time to invest all those years of experience into something new and exciting.

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GLOSSARY

Concentric: type of muscle contraction: muscle shortens during contraction

Dorsiflexion: bending the foot towards the dorsum or upper surface of the foot; opposite to plantarflexion

Eccentric: type of muscle contraction: muscle lengthens during contraction

Gait: the manner of walking

Gastrocnemius (G): superficial calf muscle; plantarflexes the ankle

Isometric: type of muscle contraction: muscle length stays the same during contraction

Leg: anatomical term for the region between the knee and ankle joints

Plantarflexion: bending the foot downwards towards the sole of the foot; opposite to dorsiflexion

Power density spectrum: depicts the amplitude² of each frequency that is present within a signal; the output from a Fourier Analysis

Soleus: deep calf muscle; plantarflexes the ankle

Supine: lying on the back, facing up

Tibialis anterior (TA): anterior shin muscle; dorsiflexes the ankle

Valgus: alignment of a joint whereby the lower, or distal, segment is angled away from the midline of the body

LIST OF ABBREVIATIONS

AS: acceleration slope

BMI: body mass index; equal to mass in kg / (height in metres)²

EMG: electromyogram

G: gastrocnemius muscle

MPF: mean power frequency; the average frequency contained within a signal

MUAP: motor unit action potential

PA: peak acceleration

PETCO₂: end tidal carbon dioxide pressure

TA: tibialis anterior muscle

TP: time to peak acceleration

VGRF: vertical ground reaction force

Chapter I

INTRODUCTION

An average long distance runner, covering 130 km/week, will impact the ground 40,000 times in 7 days (Cavanagh and Lafortune, 1980). It has been suggested that impulsive forces such as these may be responsible for hastening the degeneration of the weight-bearing joints. However, epidemiological studies of long-distance runners do not support this theory (Lane and al., 1986; Konradsen, Hansen and Søndergaard, 1990).

Impact forces in the leg are affected by many variables. Velocity, posture, age, level of fatigue and the characteristics of the foot/ground interface can all influence the amplitude and frequency of the impact shock. As the shock wave travels proximally, however, the body acts like a low pass filter, removing most of the high frequency components of the shock (Light, McLellan and Klenerman, 1980; Wosk and Voloshin, 1981; Chu, Yazdani-Ardakani, Gradisar and Askew, 1986). Consequently, most normal gait situations will lead to similar amounts of acceleration ultimately affecting the head (Wosk and Voloshin, 1981; Derrick, Hamill and Caldwell, 1998; Voloshin, Mizrahi, Verbitsky and Isakov, 1998; Ferris, Liang and Farley, 1999).

Factors that affect this filtering property may be important in our understanding of the pathogenesis of bone and joint degeneration. Poor attenuation in the leg will cause an increased shock not only in the knee and hip joints, but also proximally in the spine and head. This effect of increased shock has been shown to increase the incidence of microfractures in the tibia, as well as cartilage degeneration in the knee joint (Radin et al., 1973; Radin, Orr, Kelman, Paul and Rose, 1982; Burr, Martin, Schaffler and Radin,

1985; Farkas, Boyd, Schaffler, Radin and Burr, 1987). Healing of the microfractures causes an increase in bone stiffness (Pugh, Rose and Radin, 1973), which, in turn, decreases the attenuation properties even more.

Fatigue has been shown to decrease shock attenuation (Milgrom, 1989; Mizrahi, Voloshin, Russek, Verbitsky and Isakov, 1997; Fyhrie et al., 1998; Verbitsky, Mizrahi, Voloshin, Treiger and Isakov, 1998; Voloshin et al., 1998; Mizrahi, Verbitsky and Isakov, 2000a; Mizrahi, Verbitsky and Isakov, 2000b; Mizrahi, Verbitsky and Isakov, 2001). Most studies comparing fatigue and shock absorption, however, use global fatigue as their indicator. Unfortunately, this does not necessarily mean that the muscles of the lower leg are fatigued, nor that the different muscle compartments fatigue at the same rate (Reber, Perry and Pink, 1993; Hortobágyi, Tracy, Hamilton and Lambert, 1996; Mizrahi et al., 2000b;).

In an attempt to quantify the effect that specific muscle fatigue has on shock attenuation, other kinematic variables must be controlled. The human pendulum approach, as described by Lafortune and Lake (1995) allows the researcher to do just that. By controlling the impact velocity, surface interface and position of the leg, an impact force that closely approximates that found in normal gait is reproducible. In addition, the position of the ankle, knee and hip joints can be reasonably well controlled, which is not possible when the leg is analyzed during walking or running. This allows observation and analysis of the specific effect that localized muscle fatigue has on the attenuating properties of the shank.

Ageing of the human body brings with it two factors that could potentially affect shock attenuation: decreases in the bone density and muscle mass (Whiting and

Zernicke, 1998). Menopause is known to hasten both of these changes. The rate of bone mass loss in women is up to 10 times greater in the five years after menopause than in men of the same age (Christiansen, 1992). Thus, analyzing shock attenuation in post-menopausal women maximizes the potential changes in bone density and muscle mass, as opposed to doing a similar study with men.

1.1 Statement of Purpose

The purpose of this study was twofold:

- a) To analyze the effect that specific muscle group fatigue has on the force attenuating properties of the shank. The ankle dorsiflexors and plantarflexors were fatigued independently and compared to the non-fatigued state. Other kinematic variables such as joint angle and velocity were controlled.
- b) To analyze the effect of age on the aforementioned changes in force attenuation with muscle fatigue. Women aged 20 - 25 years were compared to those in the 50 – 60 year age group. All subjects regularly participated in some form of weight-bearing exercise at least twice a week, to ensure a reasonable degree of physical fitness and bone density (especially pertinent in the older group).

1.2 Statement of Hypotheses

Hypothesis #1: As the muscles of the leg fatigue, they will be less able to attenuate shock. Therefore, the magnitude of the peak acceleration and the rate of the increase of acceleration being measured at the tibial tubercle at the time of impact will increase. The time taken to reach peak acceleration will decrease.

Hypothesis # 2: The difference between pre-fatigue and post-fatigue shock

measurements at the tibial tubercle will be greater when the plantarflexors are fatigued than when the dorsiflexors are fatigued.

Since the mass of the plantarflexors is greater than the dorsiflexors, the degree of change in the attenuation properties should also be larger.

Hypothesis # 3: The difference between the pre-fatigue and post-fatigue shock

measurements at the tibial tuberosity will be significantly different for the older women than the younger age group.

Presumably smaller muscle mass in older women will demonstrate a smaller pre and post- fatigue difference.

Chapter II

LITERATURE REVIEW

2.1 The Effect of Impact

Radin et al. (1973) impacted the hind leg of rabbits daily to a maximum of 36 days. Synovial effusion was observed in the knee joints after 4 days of the study, with subchondral bone stiffening appearing on the 6th day. Gradual degeneration of the cartilage began around day 16, and progressed to gross cartilage changes by the end of the study. Numerous trabecular fractures were found in the cancellous bone of the impacted limbs, as well as evidence of early callus formation, in various stages of development. Cartilage changes were always preceded by subchondral bone stiffening.

Radin et al. (1982) observed the effect that walking on cement had on sheep. After 9 months, although there were no gross or histological changes in the joints, all of the sheep walked with a slight limp. Slight fibrillation of the cartilage was noticed after two and a half years, as well as early signs of osteoarthritis. The most obvious change, however, was the realignment of the trabecular structure, into a more longitudinal arrangement. This increased the strength of the cancellous bone, a direct response to the increased loading produced by cement walking.

Wolff's Law states that "the organization of bone reflects its function, or that changes in stress of the bone induce changes in its structure appropriate to support the changed load" (Pugh, et al., 1973). Consequently, bony compression in the concavity will facilitate bone hypertrophy, and tension in the convexity will encourage bone

resorption. Intermittent compression, similar to that produced in a normal gait, causes more bony hypertrophy than static compression (Chamay & Tschantz, 1972).

Trabecular fracture and repair, under normal physiological conditions, should be easily tolerated due to the high metabolic turnover of the cancellous bone (Brukner, Bennell and Matheson, 1999). Over time, as the trabeculae undergo microfracture, repair and callus formation, the cancellous bone will stiffen in the axis which is transmitting most of the impacts. The eventual alignment of the trabeculae will demonstrate an arrangement that protects them from bending stresses, which they are least able to resist (Pugh et al. 1973). Similar to the “survival of the fittest” theory, only those trabeculae that are not subjected to bending stress will be able to survive without re-fracturing. Thus, the gradual trabecular arrangement in the cancellous bone will line up with the patterns of greatest axial stress, improving the strength in these regions. Cancellous bone is, therefore, extremely anisotropic, in response to the various stresses imposed on it.

Another effect of repetitive impact loading is early vascular change in the subchondral bone. Farkas et al. (1987) demonstrated a 10% increase in vessel perimeter and bone mass after 6 weeks of impulsive loading of rabbits’ hind limbs. This increased vascularity of subchondral bone is typical of early changes seen with osteoarthritis (Mankin, 1974).

The deformation properties of subchondral bone help protect the overlying cartilage from physiological shock loads (Radin and Paul, 1970). Increased subchondral bone density is associated with an increase in bone stiffness, and directly proportional to the degree of healing (Pugh et al., 1973). Yet, focal areas of healing could potentially cause

regions of increased shock transmission to the overlying cartilage and subsequent breakdown. Thus, as the bone stiffens, its ability to protect the cartilage may diminish.

Farkas et al. (1987) also demonstrated that when impacts occur with a slower rate of loading, no changes to the bone mass or cartilage were observed. This indicates that a high rate of impact, although still within a physiological range, is much more damaging to the bone and cartilage than the same force applied at a slower rate.

Impulsive forces need not exceed physiological limits to incur damage to the body (Burr et al., 1985). Loads at the upper limit of the physiological range produced significant microdamage to the compact bone of canine forelimbs after 10,000 cycles, an amount equivalent to about 11 miles of running for a dog.

Continued microdamage can ultimately lead to complete fracture. Li, Zhang, Chen, Chen, and Wang (1985) exposed 20 rabbits to forced jumping for 2 hours daily (roughly equivalent to 360 jumps per day) for a period of 60 days. Microdamage was identifiable after 10 days in the cortical bone. After 21 days, 2 of the rabbits demonstrated an incomplete fracture of the cortex. The fractures were irregular in appearance and were a result of several smaller cracks converging. On the 50th day, one tibia demonstrated an incomplete fracture. However, most of the tibias were able to gradually adapt to the changing stress through an internal remodeling process, and thus, avoid fracture.

2.2 Normal Impact Forces During Gait

With normal walking, the forces acting upon the heel can increase to as much as 1.5 times body weight during the first 50 ms following contact (Wosk and Voloshin, 1981). In shod running and jumping, this has been known to increase to 1.8 - 10 times body weight (Dickinson, Cook and Leinhardt, 1985; Voloshin, 1988; Milgrom, 1989) with the

time taken to reach the impact peak reduced to 23 ms, or even 10 ms if running barefoot (Cavanagh and Lafortune, 1980; DeClercq, Aerts and Kunnen, 1994). Light et al. (1980) described a force of 200 N being applied over a short 4 ms period when subjects walked in shoes with a hard leather heel.

2.2.1 Force/time curve

The force/time curve for running shows two distinct peaks (Figure 1). The first peak has generally been referred to as the impact peak, and represents the initial ground reaction force as the heel strikes the ground, usually within the first 50 ms after heelstrike (Nigg, Bahlsen, Luethi & Stokes, 1987). In studies where subjects were instructed to contact the ground initially with the forefoot or midfoot, this first impact peak was nearly absent (Cavanagh & Lafortune, 1980; Farley and González, 1996). Following this, the reaction force decreases slightly, as the body's shock-absorbing structures are able to dissipate a portion of the impacting force. Examples of this would be ankle dorsiflexion, knee flexion, and compression of the weight-bearing joints, intervertebral discs and heel pad. Following this brief lowering, the ground reaction force again increases as the entire weight of the body is carried over the stance foot. This is referred to as the active peak and is generally larger than the impact peak (Liu and Nigg, 2000). The active peak averages 2.5 - 2.8 times body weight, occurs during the first 80 - 96 ms of stance (Cavanagh and Lafortune, 1980; Dickinson et al., 1985), and should occur at approximately the same time as the body's center of mass reaches its lowest point (Farley and González, 1996).

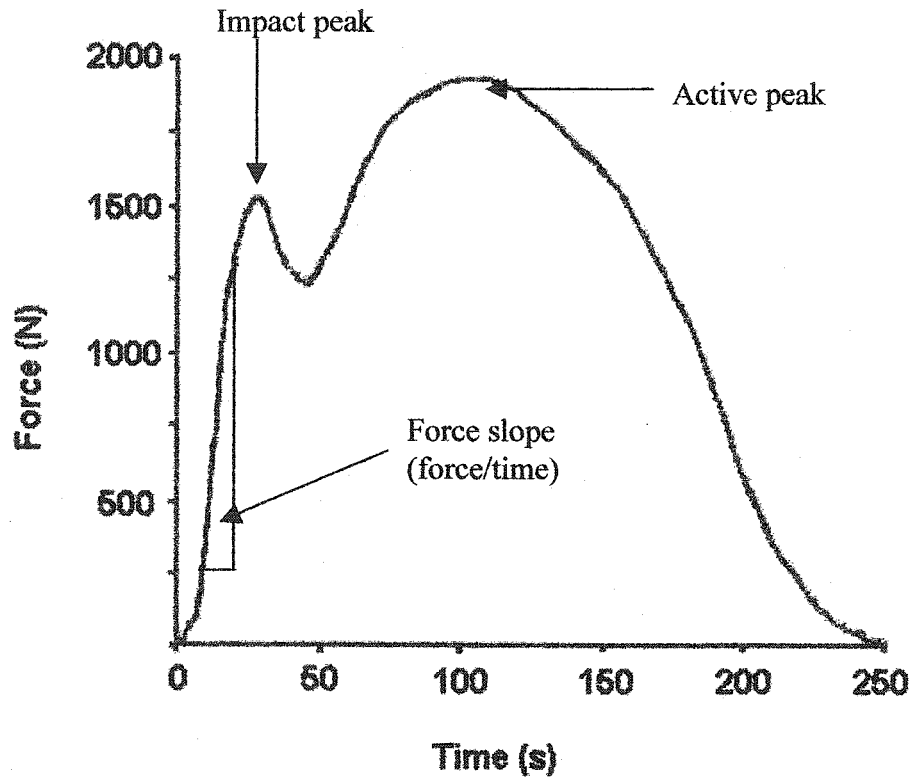


Figure 1. Typical ground reaction force curve reported for running (Modified from Lafortune & Lake, 1995)

The rate of loading, or the change in force divided by the change in time, can also be taken from the slope of the impact peak curve, using the interval that the impact force increased from 30% to 70% of its peak force (Lafortune et al., 1996). A higher rate of loading will have a different effect than a slowly applied force (Farkas et al., 1987). Rate of loading can be monitored as other variables are changed, such as shoe midsole hardness or knee flexion, to understand the effect these variables have on impact forces. For example, De Clercq et al.(1994) found the average loading rate to be 2 – 4 times higher in barefoot running compared to shod running. Faster velocities at heel-strike also result in an increase in the magnitude of the impact peak (Light et al., 1980; Farkas et al., 1987; Gerritsen, van den Bogert and Nigg, 1995; Liu and Nigg, 2000). A velocity of

0.30 m/s when walking was found to be the average by Light et al. (1980), and 1.15 m/s when running (Cavanagh et al., 1984).

2.2.2 Acceleration/frequency

The impact forces travel through the body in the form of a sinusoidal shock wave that can be measured using an accelerometer. The different frequencies found within an acceleration signal can be determined by doing a Fourier analysis, which results in a power density spectrum. This outlines the amplitude² of each frequency that is present. The mean power frequency (MPF) is the average frequency contained in a signal.

In walking, the frequency range of the impact shock wave usually falls between 10 Hz and 20 Hz. (Derrick et al., 1998; Nigg, 2001). However, frequencies as high as 75 Hz and 100 Hz were documented by Simon et al. (1981) and Paul et al. (1978) respectively, with different footwear. Lafortune, Henning and Valiant (1995) found that 95% of the power density spectrum measured by a bone mounted accelerometer in the tibia were attributed to frequencies below 35 Hz., and approximately 90% of the power fell below 20 Hz when running.

Impact frequencies can also be classified according to the phase of gait. The impact phase, associated with the high-frequency shock of heel strike, was found to be between 10 Hz and 20 Hz. (Hamill, Derrick and Holt, 1995; Derrick et al., 1998). The active phase, associated with the stance phase of gait, was found to be between 3 and 8 Hz. (Derrick et al., 1998), with the shock waves requiring approximately 10 ms to travel from the leg to the head.

Damping of the acceleration due to impact occurs during the first 75 ms after heel strike (Lafortune et al., 1995). This reflects a slightly underdamped system, which is

consistent with an earlier description of the body's ability to attenuate shock waves, published by Greene and McMahon, (1979).

Some authors will refer to the impact acceleration in terms of the number of "g's", referring to a multiple of the earth's gravity, which is 9.81m/s^2 in a downward direction. At the knee, Light et al. (1980) measured acceleration of 8 g's, using intracortical pins inserted into the tibia to anchor the accelerometer to the bone, and with varying footwear during walking. Wosk and Voloshin (1981) recorded 5 g's in barefoot subjects, using accelerometers attached to the skin overlying the tibial tubercle. Impacts as high as 30 g's at the heel during shod running were found by Chu and Yazdani-Ardakani (1986).

The healthy body is highly efficient at attenuating these forces before they reach the head. Chu et al. (1986) demonstrated on a knee specimen that an "intact, excised joint" was able to attenuate 59% of the acceleration applied to it via the tibia. Wosk and Voloshin (1981) measured the impact acceleration at the tibial tuberosity, medial femoral condyle and forehead in 39 healthy subjects during normal walking. They found that the magnitude had decreased by about 26% by the time it reached the femur, and 70% at the forehead. Light et al (1980) found that the acceleration decreased by as much as 80% during the 10 ms required to reach the head, measuring in at 0.5 g's on the forehead. Wosk and Voloshin (1981) believed that the ability to attenuate shock waves was so predictable in a healthy body, that an inability to do so could be diagnostic of degenerative changes within the skeletal system.

2.3 Anatomical Shock Absorbers

2.3.1 Bone deformation

The ability of cancellous bone to absorb forces was recognized as early as 1837 by C.A. Wistar (cited by Pugh et al., 1973). The trabeculae in the cancellous bone will absorb energy as they fracture (Radin et al., 1973). Depending on the health of the bone, there can also be varying amounts of bending, which will cause a tensile stress on the convex side and a compression stress on the concave side. In turn, compression will lead to hypertrophy of the bone on the concave side, and atrophy on the convex as per Wolff's Law (Bruker et al., 1999). Bone tissue can tolerate up to 65% more stress in compression than tension, prior to failure (Burstein, Currey, Frankel & Reilly, 1972).

2.3.2 Cartilage

Menisci and articular cartilage are visco-elastic structures, demonstrating properties such as creep and stress relaxation. Consequently, force attenuation will depend on the rate of loading and static loads will be absorbed differently than dynamic forces.

Fukuda et al. (2000) looked at the load transmission of resected porcine knee joints, and found that the compressive stress in the medial subchondral bone was between 4.0 and 5.2 times higher when the meniscus was removed. Thus, it would seem that the menisci protect the subchondral bone from extreme forces. When the articular cartilage was also removed, the subchondral bone pressures increased an additional 1.4 times.

Voloshin and Wosk (1983) found that the shock absorbing capacity of a healthy human knee is about 20% higher than one with the meniscus removed. Chu et al. (1986) found that a fully intact knee from a human cadaver was able to attenuate approximately

59% of tibial acceleration. Mechanical debridement of the articular cartilage resulted in an 11% increase in tibial acceleration, and a 23% increase in femoral acceleration. It still maintained 92% of the attenuating properties of the fully intact joint, thus indicating that the attenuating significance of articular cartilage is not large.

Similarly, Radin and Paul (1970), impacted bovine knees in various degrees of articular dissection. Their findings indicated that the cartilage and synovial tissue, while important to decrease friction, played an insignificant role in force attenuation.

In addition to protecting the underlying bone from undue stresses, cartilage also appears to protect it from resorption. Cartilage requires loading to develop in regenerating mesenchymal tissue. Lack of loading, therefore, inhibits cartilage formation, which can lead to pressure-induced fluid flow into the newly formed bone callus, causing resorption (de Rooij, Siebrecht, Tägil and Aspenberg, 2001).

2.3.3 Muscles

Radin and Paul (1970) described bone deformation and eccentric muscle activity as the major force attenuating structures in the leg. Derrick, et al. (1998) stated that, of all the passive and active structures in the leg, “muscle has the greatest potential to attenuate shock, both in terms of its ability to actively adjust the amount of shock attenuation and its large capacity to deform when stressed”. They explained that eccentric contractions of the muscles crossing the hip, knee and ankle joints were responsible for this attenuation. Up to 70% of impact force attenuation has been attributed to knee joint muscles working eccentrically (Kim, Voloshin and Johnson, 1994).

Mizrahi and Susak (1982) refer to eccentric muscle activity as the “active” mechanism for shock absorption. This is in contrast to the less effective “passive

mechanisms", such as bone and cartilage deformation, over which we have no voluntary control. Eccentric muscle activity, combined with the knee, ankle and foot joint angle changes they control, absorb forces as they lengthen, converting impact energy to strain or elastic energy. In turn, this energy is stored in the extensor muscles during deceleration, and recovered during the propulsive phase of gait (Simpson and Bates, 1990).

Nordsletten and Ekeland (1993) used muscle stimulation to produce a tetanic contraction of the triceps surae muscle in rat legs, then applied a bending moment until fracture occurred. Muscle contraction increased the ultimate bending moment of the tibia by 22%, and deflection of the tibia increased 50%. The total energy absorbed by the tibia increased 73%. Compressive forces produced with muscle contraction tend to decrease the effects of the tensile stresses associated with bending (Radin, 1986). Since bone can tolerate up to 65% more stress when tested in compression than tension (Burstein et al., 1972), the energy absorption prior to failure should also increase.

2.3.4 Intervertebral discs

The most obvious function of the intervertebral discs is that of shock absorption. The outer fibrous layers, constituting the annulus, surround a gelatinous nucleus. The layers of the annulus are obliquely oriented at an angle of 30°, with each concentric layer running in an opposite obliquity. This alignment permits absorption of the lateral stresses, much like the hoops of a barrel (Macnab, 1977). Innermost annular fibres attach to the avascular cartilaginous end plates on the vertebrae. Axial loads are transmitted via this cartilage to the nucleus, which deforms tangentially and vertically. In

turn, large pressures within the nucleus will cause the cartilaginous end plates to bulge into the adjoining vertebrae, leading to cancellous bone deformation (McGill, 2002).

2.3.5 Heel pad

Typically, being one of the first structures to contact the ground during heel strike, the heel pad is of prime importance for cushioning the impact forces. It reacts in a non-linear way during deformation: the compression modulus rising more rapidly with increasing deformation (DeClercq et al., 1994; Gefen, Megido-Ravid, and Itzchak, 2001). However, despite the excellent shock absorbing qualities of the pad, it has very little ability to store energy, thus acting more like a damper than a spring (Kim et al., 1994)

The unloaded heel pad thickness in test subjects was measured to be 14.5 mm to 15.3 mm by DeClercq et al. (1994). Güler, Berme and Simon (1998) generalized that the reported thicknesses in the literature range from 10 – 30 mm.

At heel strike, the heel pad can compress by 40% when subjects were walking (Gefen et al., 2001), and 60% when barefoot running (Light et al., 1980; Güler et al., 1998). Shod running, however, demonstrated only a 35% deformation, implying a stiffer heel pad when enclosed by a shoe. Similarly, Jorgensen and Ekstrand (1988) reported that shock absorption was increased by up to 15.5% with heel pad confinement, as the heel pad was less able to deform when confined. DeClercq et al. (1994) added that the enclosed heel resulted in a slower rate of loading, since the entire shoe and heel complex together demonstrate a larger degree of compression than the unshod heel alone. As a result, the distance over which deceleration occurred was greater, causing a longer time to decelerate, and a slower rate of loading. In turn, this would lead to a decreased impact force peak (Light et al., 1980; Liu and Nigg, 2000).

Paul et al. (1978) found that the heel pad reduced the peak dynamic force transmitted to the tibia by 20% – 28% in live rabbits. In walking humans, Gefen et al. (2001) attributed a reduced impact force transmission of 17% – 19% to the heel pad.

2.3.6 Wobbling mass

Wobbling mass refers to all of the soft tissues of the limb. Thus, in addition to the muscles, cartilage and heel pad already mentioned, it also includes fascia, nerves, vessels, body fluids, etc. In short, everything that is not bone.

Many impact models used in research represent the lower extremity with a rigid segment model. Gruber, Ruder, Denoth and Schneider (1998) claim that internal torques and forces produced by this approach are completely incorrect. It can be seen on high-speed video that the soft tissue segments of a limb will “wobble” on impact. Pain and Challis (2001), using a shank and heel pad mechanical model, found the maximum excursion of any part of the wobbling mass relative to the underlying rigid structure was found to be less than 17mm. However, Cappozzo, Catani, Leardini, Benedetti and Croce (1996) measured up to 40 mm of soft tissue movement over the lateral condyle of the femur in running subjects. It was suggested by these same authors that this “wobbling” effect plays a large role in energy dissipation. The damping co-efficient and vibration frequency of the wobbling mass can be altered by many variables, including impact touch-down velocity, type of heel (Liu and Nigg, 2000), and anticipation of impact force (Lake and Lafortune, 1998).

Liu and Nigg (2000) developed a mechanical model, which included wobbling mass segments. They were able to demonstrate that increasing the bone mass/ wobbling mass

ratio in the lower extremity caused increases in the impact force peak, but had minimal effect on the active force peak.

Pain and Challis (2001) found that adding wobbling mass to the shank portion of their mechanical model reduced the heel pad peak forces by 55%, to a level much more in keeping with in vivo observations. The shank peak forces obtained with a rigid model were over 100% greater than those found with the wobbling mass model. The same authors studied forearm soft tissue motion in human subjects and concluded that 70% of energy absorption could be attributed to the wobbling mass motion (Pain and Challis, 2002).

The amount of wobbling mass must surely have some effect on force attenuation, but, to date, there is very little in the literature comparing girth and shock absorption. Milgrom (1989), in his study of military infantry recruits found that recruits with a high lean muscle circumference of the calf, as measured by computer tomography, had fewer femoral and tibial stress fractures. Conversely, the presence of very little muscle mass in the calf was associated with a higher incidence of stress fracture, independent of the tibial bone size.

2.4 Variables That Influence Force Attenuation

Knowing which anatomical structures participate in force attenuation leads to better understanding of how the body can alter some of these variables to optimize shock absorption.

2.4.1 Joint angle and stiffness

Smith (1953) demonstrated that an 80 kg man, landing on a rigid leg from a height of 1 metre is likely to sustain severe damage to the head and neck of the femur, or even

drive it through the acetabulum in the pelvis. Obviously, landing on a rigid leg produces much higher shock forces than when the leg is allowed to flex when landing.

Most studies do not separate joint excursion and muscle lengthening, as the two are directly proportional: increased joint excursion occurs with increased muscle lengthening. In turn, as joint motion and muscle length increase, they attenuate more of the impact force. Mizrahi and Susak (1982) observed subjects dropping from a height of 0.5 metres or 1 metre, varying the amount of joint flexion they exhibited upon landing. There was a significant decrease in the impact force peak when subjects landed on the balls of their feet, as opposed to flat-footed. The addition of a “ground-roll” directly after the impact further decreased impact forces, as the range of joint flexion increased.

McMahon, Valiant and Frederick (1987) observed subjects running with a flexed posture, referred to as “Groucho running”. They found increased knee flexion decreased the vertical spring stiffness of the body to about 82% of that found in normal running. Decreasing vertical stiffness corresponds to less vertical excursion of the center of mass (Farley and González, 1996). Shock transmission from the shank to the head decreased to less than 20% of its normal value, indicating greater shock attenuation through the body with a flexed posture. However, the actual tibial shock measurements were slightly larger with Groucho running. Lafortune, Hennig and Lake (1996), using a human pendulum system, also found that an increased knee angle at impact reduced the initial impact force, but caused a higher acceleration in the shank portion.

Ankle motion in the sagittal plane has a large force attenuating ability (Mizrahi and Susak, 1982). The best shock absorption occurs when the ankle joint goes from a fully plantar-flexed position to that of full dorsiflexion, as in free fall, or the reverse order, as

seen in running. Thus, landing at heel strike with the foot in less than normal dorsiflexion would mean decreased time and motion for the foot to absorb forces prior to the active force peak. Therefore, a faster rate of loading would occur, causing an increased impact force (Light et al., 1980; Gerritsen et al., 1985; Lui and Nigg, 2000).

Farley and Morgenroth (1999) claim that the ankle is the most important joint in the leg for increasing total leg stiffness. They suggest that the moment of the ground reaction force was largest about the ankle, and therefore results in a larger net muscular and angular displacement for a given force. In addition, as the ankle is the joint closest to the ground, rotation around its axis will cause a larger rotational excursion than rotation about the knee or hip. Consequently, altered stiffness at the ankle can have a large effect on the movement of the body's centre of mass. Significant shock absorption has also been attributed to foot deformation. The foot is substantially flattened in mid-stance, thus is capable of storing elastic energy in the ligaments, tendons and muscles (Salathé, Arangio and Salathé, 1990; Kim et al., 1994). During early stance, the tibialis posterior muscle shows a high amount of EMG activity, indicating that it is primarily responsible for eccentrically controlling foot pronation (Reber et al., 1993).

Williams, McClay and Hamill (2001) conducted an injury survey of 40 runners, half of them having high arches and half reporting low arches. They found that the subjects with low arches were more prone to soft tissue injuries, knee injuries and medial injuries, whereas those with high arches were more likely to exhibit bony injuries, ankle injuries and lateral injuries. This is in keeping with the shock absorption theory of the longitudinal arch. Loss of the ability of the arch to flatten (i.e. high arches) would cause a higher shock transmission, and an increased incidence of shock-related disorders (bony

injuries). Those subjects who exhibited a chronically flattened foot would be constantly loading the ligaments, tendons and tibialis posterior muscle, causing cumulative trauma to these tissues, manifested moreso in soft tissue injuries.

James, Bates and Osternig (1978) described excessive pronation as causing too much tibial internal rotation, thus increasing the rotational stresses at the knee. Conversely, Nigg, Cole and Nachbauer (1993) suggested that subjects with a high arch would have a smaller degree of foot eversion (pronation). Since foot eversion is an integral part of gait, the tibia would tend to over-compensate for its absence, with increased internal rotation, causing increased stresses at both the knee and ankle joints.

Bates, Osternig, Mason & James (1978) suggested that hindfoot pronation, tibial internal rotation and knee flexion should all occur at the same time. Any alteration to this pattern, such as increased foot pronation, could adversely affect the forces transmitted through the leg, ultimately causing increased knee stress.

The complex kinematics of the lower extremity during normal gait are not fully understood. It would be reasonable to expect that abnormal tibial rotation might affect acceleration and shock in unforeseen ways. As the tibia rotates, different parts of the tibial plateau and femoral condyle would be abutting each other. This could potentially alter force transmission as the contact surface area and joint properties vary throughout the joint.

2.4.2 Stride length

When running speed is kept constant, but the stride length is increased, there is a larger impact load on the body (Derrick et al., 1998). However, the subjects in this study also demonstrated an increased attenuation ability, which was attributed to the active

absorption by the muscles around the ankle, hip and knee. A longer stride length will necessitate larger absolute degrees of rotation about a given center of rotation of each joint. Thus, larger joint and muscle excursions result in larger force attenuation (Mizrahi and Susak, 1982).

2.4.3 Stride frequency

As stride frequency increases, the leg, acting like a single linear spring with a mass atop, becomes stiffer. In fact, a twofold increase in stiffness was described as stride frequency increased by 65% (Farley and González, 1996). Consequently, the vertical displacement for the center of mass decreased a total of 76 % between the slowest and fastest cadences. Foot contact time with the ground also diminished by 32%. The peak vertical force decreased only slightly between the lowest and fastest frequencies (-19%). Thus, a stiffer leg segment seen with faster cadence caused less vertical displacement, shorter ground contact time, and a small decrease in the ground reaction force.

2.4.4 Velocity of running/walking

Nigg, et al. (1987) found a linear correlation between running velocity and impact force peak. As speed increased from 3 m/s – 6 m/s, the vertical touch down velocity increased by 130%, but the vertical impact force increased only 63%. This was due to the altered knee and hip angles at higher speeds, which caused a decrease in the vertical mass requiring deceleration at the moment of impact.

Simpson and Bates (1990) noted that with increased running velocity, the hip and knee extensor moments also increased, in an attempt to eccentrically dampen the impact shock and lessen the braking forces required in the direction of forward motion.

2.4.5 Bone stiffness

Increasing bone stiffness is a by-product of cancellous bone repair, subsequent to trabecular fracture (Radin et al., 1973; Farkas et al., 1987; Brukner et al., 1999). By definition, stiffness represents a material's stress divided by its strain (Whiting and Zernicke, 1998). Thus, increasing bone stiffness means that the material will deform less with impact stresses, causing a decrease in force attenuation. Conversely, osteoporotic bone, which is very soft, is an excellent shock absorber, due to the ease with which the trabeculae deform.

2.4.6 Frequency of acceleration

Due to its nature as a slightly underdamped, low pass filter, the body will tend to selectively attenuate some frequencies better than others (Hamill et al., 1995; Mizrahi et al., 2000a). When running, this results in a fairly constant level of head acceleration, regardless of the input shock.

Derrick et al. (1998) found that the high impact frequencies (10 Hz – 20 Hz) in humans were mostly attenuated prior to reaching an accelerometer attached to the forehead. Most of the frequencies reaching the head were in the 3 Hz - 8 Hz range, and were thought to represent the general up and down motion of the leg during running. Voloshin et al. (1998) measured the median frequency at the tibial tuberosity in running humans to be in the 11 Hz – 13 Hz range, but it had decreased to 7 Hz – 9 Hz by the time it reached the sacrum.

Paul et al. (1978), used rabbits to study the attenuation of impacts of differing frequencies through an intact leg. At the lower frequencies (3 Hz – 18 Hz), the authors found minimal shock attenuation, and the amount that did occur was attributed to the

heel pad more so than the bone and soft tissues. Attenuation progressed to 80% at 60 Hz, 98% at 360 Hz, and complete attenuation in the very high frequency range (500 Hz -3000 Hz).

2.4.7 Degenerative changes

In addition to bony stiffening, other factors such as pain or cartilage degeneration can affect the attenuating properties of the leg. Since a healthy knee attenuates 20% more shock than a meniscectomized knee (Voloshin and Wosk, 1983), a degenerated meniscus, with decreased height and cushioning ability, may have lost attenuating ability by a factor of less than or equal to 20%. A damaged meniscus may also tend to act like a "loose body", and score the adjacent articular cartilage, affecting its attenuation properties (Radin, Paul and Rose, 1972).

As shown by Chu et al. (1986), mechanical debridement of the articular cartilage in human knees increased tibial and femoral acceleration by 11% and 23%, respectively. Not only is stress transmitted more readily up the body, but the tissues distal to the debrided joint also experience higher levels of shock, thus joint degeneration due to excessive impulses becomes self-perpetuating.

Voloshin and Wosk (1983) found that painful knees had an attenuation ability nearly equal to that found in meniscectomized knees, even though they demonstrated no objective signs of degeneration. The same authors noted that subjects with low back pain attenuated approximately 20% less shock between the femoral condyle and sacrum than those with painful or meniscectomized knees. Interestingly, incoming forces arriving at the femur were also less in the low back pain subjects. They suggested that, as the body was less capable of attenuating forces proximal to the femur, subjects decreased the

incoming shocks by some alteration in gait patterns, to protect the proximal structures from abnormally large shocks.

2.4.8 Shoe and surface stiffness

Much research has been done in an attempt to quantify the benefits of different styles of shoes, and how the body adapts to ground surfaces of varying stiffness. Light et al. (1980) looked at the skeletal transients in walking with varying heel qualities. They found that, when subjects wearing hard leather heels walked across the force plate, not only did the impact peak force increase, but the rate of loading also increased significantly (to about 200 N in 4 ms). Liu and Nigg (2000), using a mechanical model, also found that the impact force peak and rate of loading increased in a hard shoe compared to one with a soft sole. However, Wright et al. (1998) found that, although rates of loading were higher with a rigid soled shoe, the peak impact force did not change. Nigg et al. (1987), observed 14 subjects running with differing midsole stiffness. They found no significant increase in the impact force peak or rate of loading with a stiffer shoe. Minimal changes were noted in the actual bone-on-bone contact forces of the subtalar and ankle joints as midsole thickness was varied by Cole, Nigg, Fick and Morlock (1995).

It has been suggested that some passive, internal change, such as altered muscle kinematics may be responsible for a lack of force increase with rigid soles (Wright et al., 1998). When wearing a shoe with more rigid soles, runners may use various strategies to maintain a constant impact force, such as slowing of rearfoot pronation, changing knee and ankle joint angles, altering joint stiffness and/or muscle tuning (Nigg et al., 1987; Cole et al., 1995, Nigg and Liu, 1999).

When surface stiffness is included as part of the total stiffness analysis, researchers have found that the combination of leg stiffness and surface stiffness remains reasonably constant, allowing consistency in the ground contact time and center of mass kinematics (Ferris and Farley, 1997; Ferris, Louie and Farley, 1998). For example, as runners knowingly encountered a decrease in surface compression from 6 cm to 0.25 cm., they decreased their leg stiffness by 29%, thus maintaining a constant vertical displacement of the center of mass (Ferris, Liang & Farley, 1999). Subjects were actually able to adjust stiffness for the first step on the new surface, indicating that the changes were not a result of proprioceptive feedback after touchdown, but were an anticipatory “tuning” of the muscles.

2.4.9 Resonance

The force of impact acceleration at heel strike during running falls between 10 Hz and 20 Hz (Derrick et al., 1998; Nigg, 2001). The natural frequency of bones tends to be high, in the range of 200 Hz - 900Hz, whereas that of the soft tissues lies between 5 Hz and 65 Hz, thus similar to the input frequencies. Yet, experimental observations demonstrate that soft tissue vibrations are short, heavily damped, and typically below 5% of the initial amplitude after 2 oscillations (Nigg and Wakeling, 2001). It was suggested that muscles are “tuned” prior to heel strike to a level of tension that minimizes soft tissue resonance, based on previous experience of input frequency, amplitude and time. Altering muscle tension changes the natural frequency of the soft tissues (Wakeling and Nigg, 2001a; Wakeling and Nigg, 2001b). This may partially explain the observations by Nigg et al. (1987) and Wright et al. (1998), that changing the

foot/ground interface stiffness does not affect the peak impact force, despite changes in rate of loading.

2.4.10 Muscle fatigue

The most consistent finding with general body fatigue is that of increasing shock transmission through the leg (Milgrom, 1989; Mizrahi et al., 1997; Fyhrie et al., 1998; Verbitsky et al., 1998; Voloshin et al., 1998; Mizrahi et al., 2000a; Mizrahi et al., 2000b; Mizrahi et al., 2001). Milgrom (1989) described an increase of 20 % – 30% in tibial acceleration amplitude following a 24 km march in military recruits. This tendency is less in older people, presumably because there is less muscle mass to fatigue (Milgrom, 1989; Fyhrie et al., 1998).

Voloshin et al. (1998) noticed a shift towards the higher frequencies of the tibial acceleration with fatigue. However, the median frequency measured at the sacrum did not show similar changes, and its acceleration magnitude increase was less than that of the tibia, indicating that the body is still tending to protect the proximal structures by proportionally better attenuation of the high frequencies as the shock travels up the body.

Verbitsky et al. (1998) described how a decreasing stride rate with fatigue was directly correlated with an increase in the shock wave amplitude. This differs from research done by Nigg et al. (1987), who found increasing stride rate proportional to increasing impact force. The authors suggested that, since the stride rate was forced upon the subjects in the Nigg et al. (1987) paper, normal running biomechanics would be altered, which could affect the results.

Pinniger, Steele and Groeller (2000) looked specifically at the effect of isolated hamstring fatigue on running mechanics. During the swing phase, they found decreased

hip and knee flexion, as well as decreased angular velocity of the leg prior to touch-down. In addition, rectus femoris stopped working sooner during the late swing phase, thus decreasing eccentric stress on the hamstrings.

Christina et al. (2001) fatigued the dorsiflexors and the invertors of the foot, in isolation from each other, then examined changes in ankle motion and vertical ground reaction force (VGRF). Dorsiflexion fatigue resulted in an increase in the loading rate associated with heel strike, but the peak VGRF remained the same. There were also significant decreases in the amount of dorsiflexion at contact. Fatigue of the invertors caused a decrease in the VGRF and a more pronated position of the foot at heel strike.

Loading imbalances in the tibia associated with general body fatigue were described by Mizrahi, et al. (2000b). EMG analysis of tibialis anterior demonstrated a significant decrease in its mean power frequency (MPF), whereas that of the gastrocnemius muscle significantly increased. Since decreasing MPF is associated with muscle fatigue (Petrofsky, Glaser and Phillips, 1982), it appears that in running, the pre-tibial muscles fatigue much more readily than the calf muscles. This would tend to increase the tibial posterior compression stress, while increasing the tensile stress anteriorly. When this tendency is combined with the increase in acceleration demonstrated with fatigue, the tibia may become more susceptible to stress fractures.

Increases of 26 – 35% in peak principle strain in the tibia associated with fatigue were demonstrated by Yoshikawa et al. (1994). In addition to changes in the total amount of shear, compressive and tensile strain, the authors noted a significant change in the strain distribution, most notably in the posterior cortex of the bone. Due to the anisotropic quality of bone, this could greatly affect the tibia's ability to absorb shocks.

The Type I (slow twitch) muscles are mainly responsible for converting bending stress to compression stress (Radin, 1986). As they fatigue, this conversion may start to fail, leaving the tibia more susceptible to damaging tensile stresses.

The effects of fatigue on bony deformation of the second metatarsal bone was studied by Arndt, Ekenman, Westblad & Lundberg (2002). They found that the bone demonstrated an increased dorsal compressive strain, and decreasing tensile strain, post-fatigue. More importantly, in the non-fatigued state, the bone experienced periods of unloading between peaks of compression. Post-fatigue, not only did the compressive forces increase, but the periods of unloading decreased, potentially causing an increase in cumulative microtrauma to the bone.

Reber et al. (1993) explained the predisposition of tibialis anterior to early fatigue by noting that it has a much different firing pattern than other muscles of the leg. It sustains a higher level of activity throughout all phases of running, whereas peroneus brevis, gastrocnemius and soleus all cycle through times of high and low activity. Tibialis posterior appears to control pronation during early stance, but has little activity at other times. As fatigue progresses, increasing pronation occurs, thus putting more stress on tibialis posterior (Thorwesten, Fromme, Winkelmann, Reer and Jerosch, 1997).

Fatigue appears to affect eccentric muscle activity less than concentric or isometric. Hortobágyi et al. (1996) compared plantarflexion fatigue, using a dynamometer, and found that after 50 repetitions, isometric strength decreased by 41%, concentric decreased by 32%, yet eccentric muscle force did not change significantly. This could affect the co-ordination of joint movements, and would help in determining the type of contraction that would most likely facilitate fatigue effects in an experiment.

Gollhofer, Komi, Miyashita and Aura (1987) described how the transfer of mechanical energy between the concentric and eccentric phases of muscle activity was reduced with fatigue, and could also affect co-ordination. With repeated eccentric/concentric cyclic exercises on weight-bearing arms, fatigue was found to decrease the angular displacement of the elbow. The rate and degree of elbow flexion (eccentric) increased, yet the rate and amount of elbow extension (concentric) decreased. This implies that the transfer of energy between the two phases is reduced. In that this transfer is one of the major ways the body has of attenuating shock (Mizrahi and Susak, 1982; Derrick et al., 1998), a reduction in efficiency would promote increased shock propagation.

General body fatigue of the runner may lead to changes in co-ordination and activation levels of muscles prior to touch-down (Dickinson et al., 1985; Gollhofer et al., 1987; Bobbert, Yeadon and Nigg, 1992; Mizrahi et al., 1997; Fyhrie et al., 1998). This, in turn, would affect impact dynamics. It appears that a deterioration in the complex, synergistic muscle recruitment patterns affects running kinematics more so than fatigue of any specific muscle. Consequently, measuring total body fatigue with a method such as end-tidal carbon dioxide pressure (PETCO₂) will not be measuring the same fatigue effects as specific muscle measurements, such as EMG amplitude and frequency.

2.5 Fatigue and Wobbling Mass

Although much research has been done on force attenuation and fatigue, very little has been done to quantify the amount of force which is specifically absorbed by the wobbling mass in a living, human subject, and how this is affected by fatigue.

2.5.1 Measuring fatigue

Using end-tidal CO₂ pressure (PETCO₂) as a measure of fatigue assumes that global, cardio-vascular fatigue will correspond to local muscle fatigue. As shown by Mizrahi et al. (2000b), muscles fatigue at differing rates, with dorsiflexors tiring before gastrocnemius. Using PETCO₂ as the fatigue indicator, it would be wrong to assume that any change in shock attenuation could be attributed to fatigue in the gastrocnemius muscle. Yet, the size of this muscle suggests that fatiguing it would have a large effect on shock absorption. To calculate the specific effect of fatigue on the attenuation properties of the muscles, researchers must ensure that the muscle is, indeed, fatigued.

Electromyographic analysis of muscles can provide information that assists in the objective measurement of muscle fatigue. Since fatigue is a dynamic process, the muscle force generating ability decreases gradually. Concurrently, the mean power frequency (the average frequency contained in the electromyographic signal) decreases, and the intensity of the signal increases. The latter, however, will vary with force generation, thus the mean power frequency (MPF) is a better indicator of fatigue. Shorter muscle length will provide a larger fatigue-induced decrease in the MPF (Potvin, 1997).

Although the literature at this time does not specify a recommended amount of MPF change to target, decreases of 11% and 16% in the medial and lateral gastrocnemius muscles, respectively, have been described with fatiguing isometric exercise (Ament, Bonga, Hof and Verkerke, 1993).

2.5.2 Changes in joint kinematics

As humans run or walk to a point of fatigue, many biomechanical changes will occur (e.g. joint angles and muscle co-ordination). If the dorsiflexors are the first muscles to

fatigue in running (Mizrahi et al., 2000b), an increased rate of impact loading, as well as a lesser amount of dorsiflexion at heelstrike would be expected (Christina et al., 2001). Both of these variables will affect force attenuation, yet this does not indicate that the wobbling mass of the leg is actually absorbing less shock.

Whether the fatigue develops first in the proximal or distal muscles, the position of the entire lower quadrant will be affected. For example, increased foot pronation with fatigue (Fromme et al., 1997) will increase the medial torsion at the sub-talar and ankle joints and cause an increase in the amount of tibial internal rotation (James et al., 1978; Nigg et al., 1993; Bellchamber, T.L. and van den Bogert, 2000). A pronated foot has also been associated with an increased valgus effect at the knee, referred to as an increased Q-angle (Williams et al., 2001a). In turn, this can cause malalignment of the patellofemoral joint, knee joint stresses, and increased internal rotation of the femur in an attempt to reduce the rotational stresses at the knee. Increased peak knee flexion angle has also been associated with runners who have pronated feet (Williams, McClay, Hamill and Buchanan, 2001b). This would require an increase in the quadriceps muscle force to eccentrically control the effect gravity has to further flex the knee.

Conversely, weakness in the hip external rotators is a muscle imbalance problem frequently seen in a rehabilitation setting (Sahrmann, 1966). This results in a chronically internally rotated femur, increased Q-angle, internal rotation of the tibia, and eventually, a flattened or pronated foot. Weak external rotators may be the first muscle to fatigue in a running situation, causing a cascade of biomechanical changes in the limb, all of which could affect attenuating properties. These potential changes must be controlled in order to quantify the specific effect of muscle fatigue on the soft tissue attenuation properties.

2.6 Human Pendulum

This lack of control over initial impact conditions during human locomotion has limited our understanding of the mechanical factors involved at impact (Bobbert et al., 1992). Lafortune and Lake (1995) first described the human pendulum approach of loading the lower extremity, in an attempt to control these conditions (Figure 2).

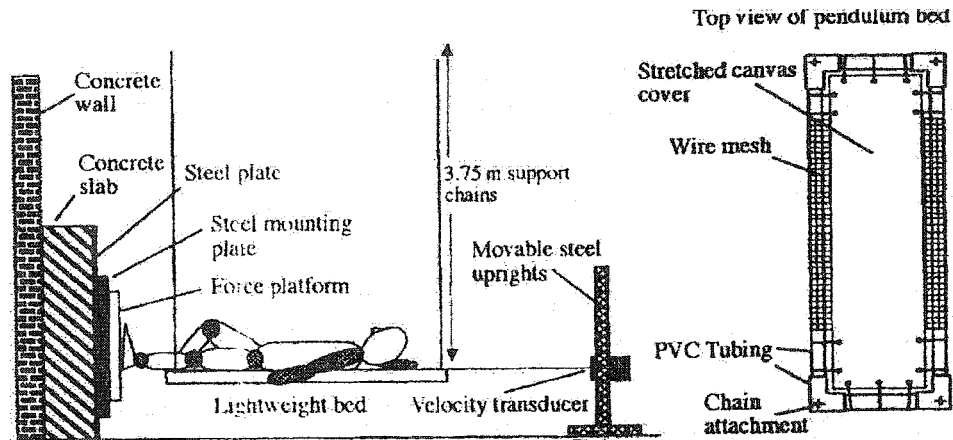


Figure 2. Human pendulum apparatus (Lafortune & Lake, 1995)

Using this technique, the subject is positioned supine on a reinforced canvas platform, which is suspended from the ceiling. The platform is pulled back, away from the force plate. Upon release, the platform and subject swing caudally. The heel impacts a vertically mounted force plate when the apparatus reaches the lowest position of the arc of swing. This approach allows forces to be generated that mimic those of walking, without the risk of the lower limb position being altered by other factors, such as muscle tuning or fatigue.

This method generates impact measurements that are more similar to those experienced in vivo, compared to a missile type impact. For example, Munro, Miller and Fuglevand (1987) described initial impact forces in running as being in the region of

1270 N in 30 ms, with total ground contact time being 247 ms. Using a missile impact, maximum forces of only 850 N could be produced without the subjects experiencing pain. The peaks occurred in just over 15 ms, and the entire contact phase lasted less than 40 ms (Cavanagh, Valiant, & Misevich, 1984; Aerts & DeClercq, 1993). Using a human pendulum technique, an impact force peak was achieved in 36 ms, with total impact time averaging 200 ms. Although not measured in Newtons, the peak force attained was 2.0 times that of body weight, which compared favourably with a range of 1.8 – 2.1 times body weight in the literature (Munro et al., 1987; Nigg et al., 1987).

Overall, the human pendulum approach was able to produce time and frequency domain variables that were within one standard deviation of values obtained from running at a speed between 3.25 and 3.75 m/s. The one exception was that the peak shank acceleration occurred slightly sooner with the pendulum. It was suggested that this occurred because the knee was held in full extension. However, in 1996, when Lafortune and Lake added a leg suspension system to compare varying knee angles with pendulum impact force, the results demonstrated an even shorter time to peak impact force with increasing knee angles. This might be explained as altered muscle tuning in gait compared to the pendulum, as well as the absence of other shock absorbing properties, such as foot pronation and ankle dorsiflexion.

The human pendulum approach to analyzing the effect of impact on the leg allows the observer to remove many confounding variables that could affect the results. Positional changes in the joints and muscles associated with fatigue can be controlled, facilitating a more specific analysis of the effect fatigue has on the shock attenuating properties of the wobbling mass.

2.7 Measuring Acceleration

In the literature, acceleration in the proximal shin has been measured using either skin-mounted transducers (Wosk and Voloshin, 1981; Hamill et al., 1995; Mizrahi et al., 1997; Derrick et al., 1998; Lafortune et al., 1998; Verbitsky et al., 1998; Voloshin et al., 1998; Mizrahi et al., 2000a and b) or transducers affixed to the bone via metal pins (Light et al., 1980; Lafortune et al., 1995). Although the second method provides a more direct method of measurement, it requires surgical insertion of the pin into the cortex of the bone. Consequently, there is a moderate surgical risk associated with it. The use of low mass surface accelerometers, preloaded onto the skin surface with elastic straps, has been shown to successfully measure bone vibration in a non-invasive manner (Kim, Voloshin, Johnson and Simkin, 1993; Valiant, McMahon, and Frederick, 1987).

Chapter III

METHODOLOGY

3.1 Subjects

Two groups of volunteer female subjects participated in the study, with 12 women in each group. Women between the ages of 20 and 25 years were recruited from the university population (Appendix 1). The second group consisted of women between 50 and 60 years of age. This older group was recruited by word of mouth amongst acquaintances, as well as at a local running club. Consequently, six out of the 12 older women were distance runners. The subjects' average mass, height and body mass index ($BMI = \text{mass}/\text{height}^2$) are shown in Table 1.

Table 1. Subject demographics

Age group (years)	number	Age (years)	Mass (kg)	Height (m)	BMI (wt/ht ²)
20 - 25	12	23.0 \pm 1.5	60.3 \pm 10.2	1.647 \pm 0.046	22.2 \pm 3.1
50 - 60	12	53.4 \pm 2.5	63.8 \pm 7.8	1.626 \pm 0.048	24.4 \pm 2.4
total	24	n/a	62.2 \pm 9.0	1.636 \pm 0.047	23.3 \pm 3.0

Subjects were screened by a registered physiotherapist for any pertinent history, evidence of poor bone mineral density, or pain in the knee, ankle, hip or low back. At least 5 subjects were rejected prior to the initial visit due to the above reasons. Two women from the older group did not return for the second impact session; one due to illness and the other due to attrition. In the younger group, one subject's data were discarded as she had a very limited amount of dorsiflexion, which impaired her ability to impact the force plate with her heel.

All subjects were given a consent form that described the experiment prior to proceeding (Appendix 2). The completion of this form was accepted as an indication that the person had a good understanding of the requirements and risks of the experiment. The research was approved by the Research Ethics Board of the University of Windsor (Appendix 3).

3.2 Experimental Apparatus

The apparatus used for the human pendulum is based on that used by Lafortune and Lake (1995) (Figure 2, see page 32).

3.2.1 Suspension/impact apparatus

The metal frame was constructed of 2.5 cm diameter iron pipe, screwed into right angle iron joiners at each corner. The outside measurement of the frame was 52 cm by 190 cm. A rectangle of firm canvas was lashed to the pipes, providing a reasonably firm support for a supine subject. This frame was suspended at each of the four corners from ceiling supports that are 280 cm above the platform. Airplane wire, tensioners, and miscellaneous joiners produced a suspension system that was light, adjustable, and imparted minimal friction to the pendulum. The total mass of the frame and suspension apparatus was 13 kg (Figure 3).



Figure 3. Human pendulum apparatus used for this study

Each subject was strapped down to the frame with 2 straps made from 5 cm wide webbing, to prevent unwanted horizontal movement. One strap was secured around the pelvis, while an additional strap was applied just proximal to the knee, to also prevent knee flexion. The unshod, dominant leg protruded over the end of the bed such that impact occurred when the bed was at the lowest point, and traveling tangentially towards the force plate, in a horizontal direction. Although an initial knee angle of 30° flexion is in keeping with that found in running humans at heelstrike (Wakeling, von Tscharner, Nigg and Stergiou, 2001), it was felt that this would be difficult to maintain during fatiguing exercises. Therefore, the subject was asked to maintain the leg in full extension throughout the experiment.

The platform was pulled backwards by a rope that traveled through a wall-mounted pulley. A metal barrier was used to mark the horizontal distance the pendulum was to be retracted, to maximize impact consistency with repeated trials. A velocity of 1.0 - 1.15 m/s was targeted, as this has been described as the impact velocity when running at 3.6 m/s (Cavanagh et al., 1984), as well as a peak impact force approximating 1.8 – 2.8 X body weight (Cavanagh and Lafortune, 1990; Dickinson et al., 1985).

3.2.2 Data acquisition

Impact forces were measured by a force platform (AMTI OR6-5-1) mounted vertically (Figures 4, 5). This was bolted to a 2 cm thick steel plate (56 cm X 61 cm) as well as to the mounting frame. The frame was fabricated from 5 cm X 5 cm tubular steel, which was, in turn, bolted to the concrete floor.

Linear velocity of the pendulum was measured by means of a cable

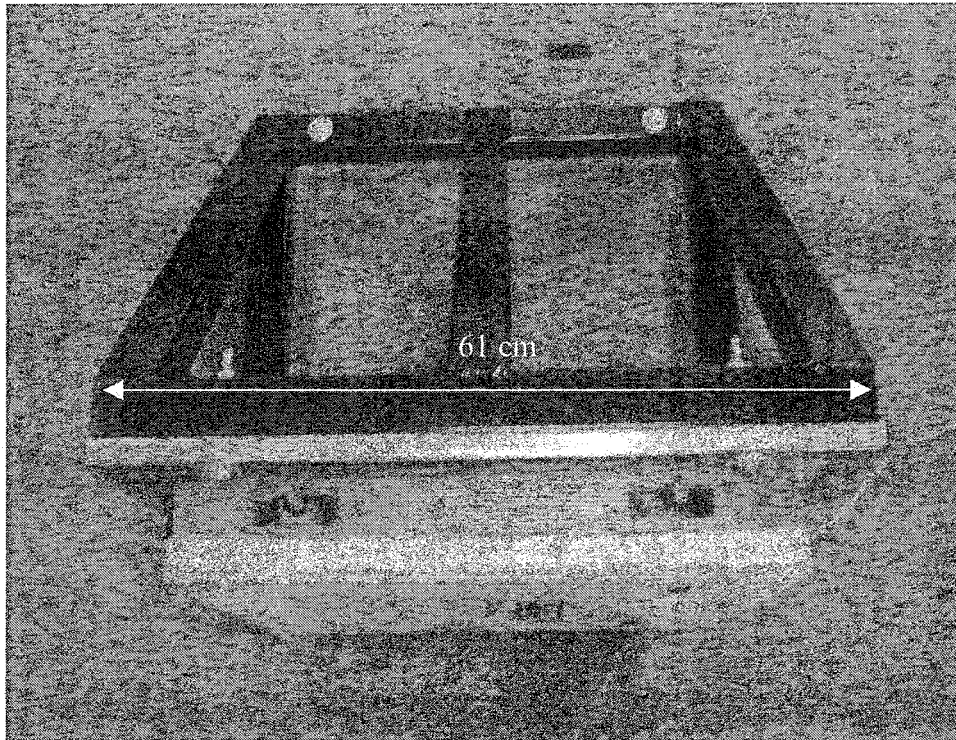


Figure 4. Force platform mounting frame, top view

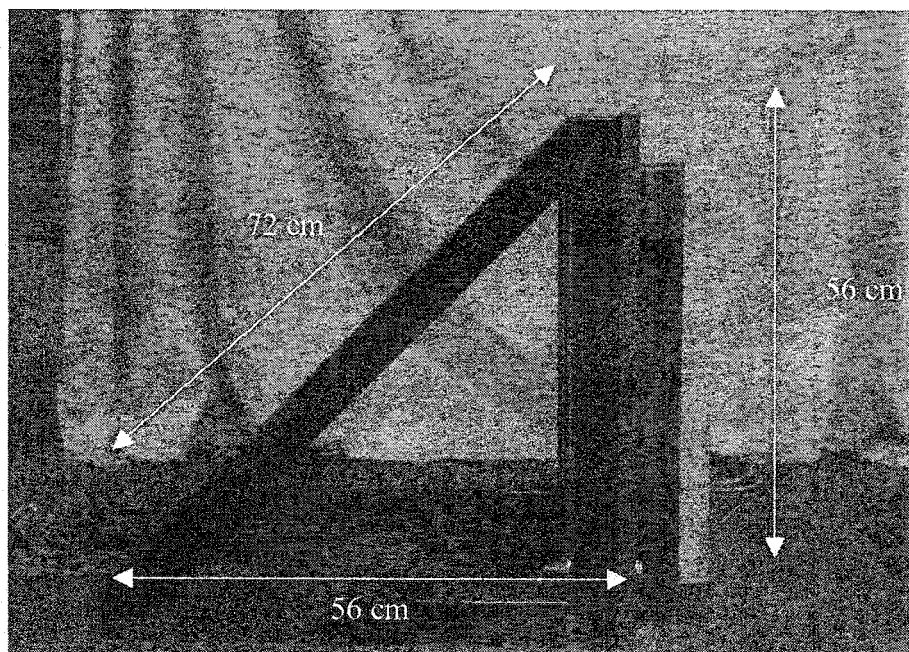


Figure 5. Force platform mounting frame, side view

velocity/displacement transducer (Celesco DV301). This was firmly attached to the wall and the metal frame, such that cable displacement occurred as the pendulum swung towards the force plate.

A miniature accelerometer (Entran EGA-25-/C) was used to measure uni-axial linear acceleration of the proximal shank. It was mounted on a small piece of balsa wood (approximately 30 mm X 18 mm X 2 mm) that was attached to the medial aspect of the tibial tuberosity on the tibia, approximately 2.5 cm distal to the knee joint line. The combined accelerometer and balsa wood weighed 1.623 grams. Acrylic glue and Velcro™/elastic straps were used to secure its attachment, with the sensitive axis visually aligned parallel to the longitudinal axis of the tibial shaft. The strap preloaded the accelerometer with a force of approximately 45 Newtons in a direction normal to the shaft of the tibia. The acceleration signals were amplified close to the source (Entran IMV-15/15/5VM-1W/L6F). Figure 6 depicts the three dependent variables that were analyzed: peak acceleration (PA), the time taken to reach peak acceleration (TP), and maximum acceleration slope, measured between 30% and 70% of the total rise (AS), as per previous works in the literature (Lafortune et al., 1996).

EMG analysis was done on the tibialis anterior and lateral gastrocnemius muscles, due to their superficial proximity and ease of signal acquisition, using Kendall Meditrace 130 disposable 10 mm circular Ag/AgCl bipolar surface electrodes. After hair removal and cleansing of the skin with isopropyl wipes, two electrodes were placed over the muscle belly of either the tibialis anterior or lateral gastrocnemius, depending on which muscle was being tested in that session. The intra-electrode distance was approximately

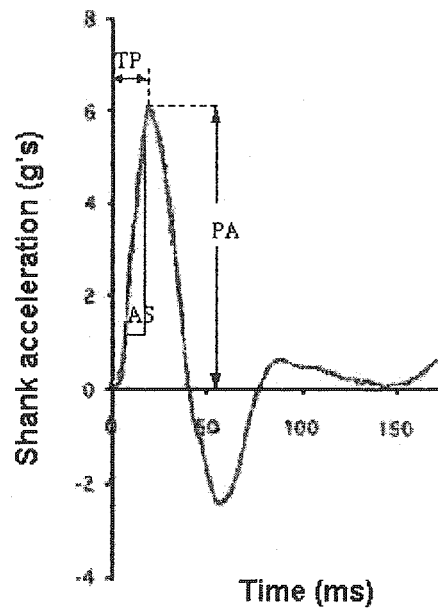


Figure 6. Acceleration/time graph, depicting the 3 dependant variables: peak acceleration (PA), time to peak acceleration (TP) and acceleration slope (AS) (modified from Lafortune & Lake, 1995)

30 mm, centre to centre. A ground electrode was placed over the patella. The frequency and amplitude of the EMG signal was monitored during the fatiguing exercise, to ascertain the occurrence of fatigue. A mean power frequency decrease of approximately 15% was targeted to indicate fatigue (Ament et al., 1993), as well as visual cues such as muscle trembling and changing ankle joint angle.

All data were collected and stored using Labview software, at a sampling rate of 4000 Hz. Acceleration, velocity, force and %MVC data collection was triggered by impact, with the 50 ms before and after impact being logged for analysis. All of the data were collected in a double buffered system: one buffer collecting data for 50 ms prior to and after peak impact, and the rest of the data being stored in a collector buffer.

3.3 Testing Sessions

Data collection took place during two separate sessions. The first session involved fatiguing the anterior compartment of the shank (the dorsiflexors), while the second induced posterior compartment fatigue of the plantarflexors. To allow the muscles of the leg adequate time to recover from the fatiguing effects of the first session, the second session took place at least 7 days later. Even though different muscles were being fatigued, it was felt that residual effects from the fatigued anterior compartment might affect the impact attenuation in the second session if the two were carried out closer in time.

3.3.1 Session 1

The first session began with the subject reading the “Consent to Participate” form (Appendix 2). Any questions that arose were answered and once the examiner believed the subject had a good understanding of the procedure and risks involved, they were asked to sign the “Rights of Research Subjects” form (Appendix 2). Subjects were asked about any history of pain in the back or leg, or low bone mineral density. If this was present, a brief musculoskeletal assessment of the lower limbs and back took place, to ensure no pathology was present which would preclude their participation in the study. Subjects were asked which leg they would kick a soccer ball with, to determine leg dominance. Maximum voluntary contractions (MVC) of the two muscle groups were then elicited with concurrent EMG analysis. To obtain the dorsiflexors’ MVC, subjects were encouraged to pull their foot into full dorsiflexion, while heavy resistance was applied manually by one of the experimenters, until the isometric hold was broken. This was repeated three times, and the highest amplitude was accepted as the MVC.



Figure 7. Plantarflexor resistance system for obtaining MVC

Plantarflexor MVCs were collected in standing, due to the great strength of the gastrocnemius muscle (see Figure 7). Two pieces of non-flexible webbing were fed through eyebolts on a horizontal platform, and led over the subject's shoulders. It was tightened to a length that maintained the upright subject in slight knee and ankle flexion, and then secured at this length. Subjects were encouraged to push hard against this system, attempting to plantarflex both ankles maximally against the resistance. Again, this was repeated three times and the highest amplitude of the EMG signal was accepted as the MVC.

The subject then lay supine on the suspended pendulum, with the dominant leg projecting over the end, and the opposite hip and knee bent to allow the subject to hold the non-dominant knee against her chest. This was done to prevent the non-dominant leg or arms from pushing against any part of the suspension frame during impact. The subject aligned herself in a position that placed the bare heel in light contact with the force plate when the suspension apparatus was at rest. There were two different securing positions of the vertical suspension wires on the ceiling, which allowed the entire platform to be moved approximately 5 cm further or closer to the force plate, allowing for individual variances in tibial length. The two webbing straps were then secured: one around the pelvis and one proximal to the knee. At this time, the accelerometer was also glued and strapped to the subject's proximal leg, over the medial aspect of the tibial tubercle. The medial and lateral knee joint line was palpated and

marked with a permanent marker, as well as the position of the accelerometer. Subjects were asked to maintain these markings between test sessions, to ensure similar positioning was obtained. This facilitated continuity in the subject's position relative to the end of the bed, both between and within sessions. After application of the accelerometer (as above), the pendulum was pulled back approximately 0.55 ± 0.03 metres in order to result in an impact forces and velocities similar to those found previously in running (Lafortune and Lake, 1995). The pendulum was then released, and the ensuing impact force and velocity were analyzed, to ensure outcomes that were within the parameters set as acceptable for this experiment: 1 – 1.15 m/s for velocity, and an impact force of 1.8 – 2.8 X body weight. When the acceptable distance was found, impacts were repeated 3 times, and the mean score calculated. This provided the pre-fatigue impact data for the dorsiflexors.

Subsequently, the pendulum was pulled back the above distance and secured in this position, using a one-way cleat to secure the rope, as shown in Figure 8. A resistance apparatus was then lowered into the space between the force plate and the pendulum (Figure 9). The distal portion of the dominant foot was slid under a stiff rubber bung

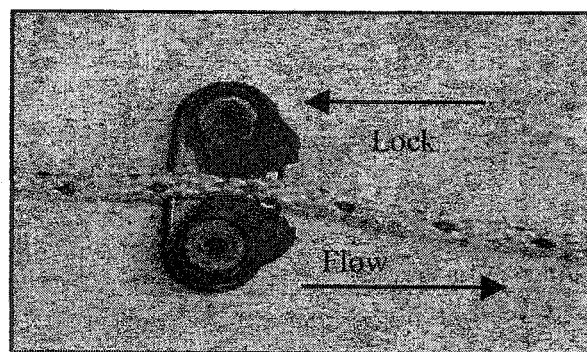


Figure 8. One-way cleat and rope used to secure the pendulum, allowing for quick release of the system.

cord, and the subject instructed to dorsiflex the foot isometrically and maximally until fatigued. Subjects were encouraged to sustain the contraction as long as possible, ideally at a level approximating 50% of their MVC. This output measure was visible to the

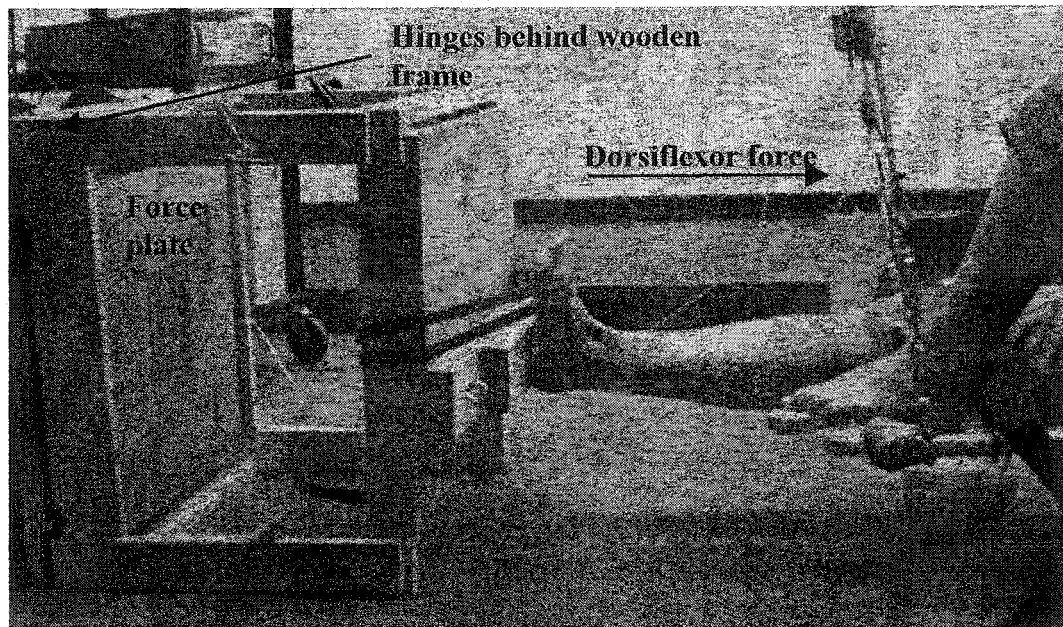


Figure 9. Resistance apparatus applied to the force platform frame. The hinges allowed for easy removal once dorsiflexor fatigue was achieved.

experimenter, thus the subject could be encouraged to pull harder or ease off, as required to maintain 50% MVC. The mean power frequency of the EMG signal was simultaneously monitored, watching for signs of significant fatigue (minimum of 15% decrease in the shift in the mean power frequency) (Ament et al., 1993). Other gross signs such as trembling of the leg or changing ankle joint angle were also taken as signs of increasing fatigue. When significant fatigue had occurred, the apparatus was quickly lifted out of the way and the securing rope released. Impact occurred an average of 20 seconds after the resistance had been removed. Subjects were asked to maintain

the isometric hold until the pendulum was ready for release. At that time, they were encouraged to focus on impacting the force plate with the heel, and avoid “slapping” the force plate with the forefoot. Three impacts were repeated quickly, with the force and velocity outcomes being checked briefly after each impact to ensure consistency. The dorsiflexors were then fatigued again, using the same protocol as the first time, followed by another 3 impacts. Thus, the protocol for a given session consisted of three pre-fatigue impacts, fatiguing of the dorsiflexors, and 3 post-fatigue impacts. Isometric fatigue and three post-fatigue impacts were then repeated.

This second series of fatigue/impacts was felt necessary to ensure fatigue was maintained, yet allow a minimum of 6 post-fatigue impacts from which to observe outcomes. Data from three trials were then chosen from the six, based upon the impact velocity and impact force. Trials were used which fell within the target parameters and had a clear impact peak (ie no foot slap). If all six trials met both of these requirements, then the three which had the most similar peak impact force were used for analysis of the dependant variables.

3.3.2 Session 2

The procedure on the second day began with collecting MVCs of the tibialis anterior and gastrocnemius muscles, as was done on Session 1. The supine pre-fatigue impacts described in Session 1 were also repeated, with the dominant leg again being used for testing. Repeating this impact allowed a check to see if any significant day-to-day changes in non-fatigued attenuation properties occurred that would affect the comparison of plantarflexor and dorsiflexor fatigue.

To fatigue the plantarflexors, heavy webbing was used to secure the pendulum to the force plate apparatus. The tension in the securing system was such that it allowed approximately 5 cm of motion between the bed and the apparatus, facilitating isometric activity in a position close to the inner range of the gastrocnemius muscle (i.e. approximately 45 degrees of plantarflexion). Again, the subject was encouraged to continue with this isometric activity at a level approximating 50% of their MVC. The observer watched for a decrease in the EMG mean power frequency, as well as leg trembling or changing ankle joint angle.

Once the observer was satisfied that the plantarflexor muscles were fatigued, the web securing system was quickly removed. The pendulum was pulled back to the pre-determined distance and released for impact. Again, 3 sets of post-fatigue impact data were collected; a second session of fatigue took place, followed by another 3 impacts.

3.4 Free Body Diagram

The free body diagram of the forces involved in the foot and shank are shown in Figure 10. As can be seen in the diagram, the quadriceps exerts forces in the positive x and y axis directions, maintaining the knee in extension. Dorsiflexion of the foot is produced by the tibialis anterior force being greater in the x direction than is that of gastrocnemius, causing a dorsiflexion moment at the ankle joint. At impact, the force plate will exert a large positive force along the x-axis posterior to the ankle joint, causing a plantarflexion moment at the ankle, which must be resisted by the dorsiflexors.

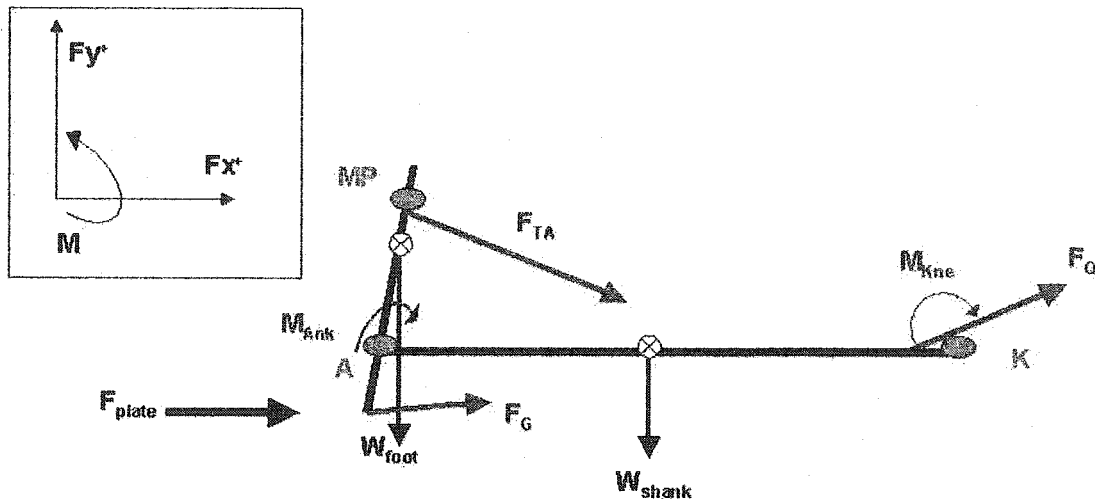


Figure 10. Free body diagram of the foot and leg at the time of impact (A = ankle joint; F_G = Gastrocnemius Force; F_Q = Quadriceps Force; F_{TA} = Tibialis Anterior Force; F_{plate} = impact force from plate; K = knee joint; M_{Ank} = Moment about the ankle; M_{kne} = moment about the knee; MP = Metatarsal-phalangeal joints; W_{foot} = Weight of foot; W_{shank} = Weight of shank)

3.5 Statistical Analysis

There were 3 independent variables in this study: one between subject variable (age) and two within subject variables (fatigue state and muscle group). This produced 8 experimental conditions (Table 2). The dependant variables included peak tibial impact acceleration, time to peak, and the rate of the rise of acceleration. The impact forces and velocity, although ideally standardized, were observed for variability within and between groups, as was the %MVC of both muscle groups at the time of impact.

Table 2: Study design

Between Subject Variable (Age)	Within Subject Variables			
	dorsiflexors		plantarflexors	
	Prefatigue	Postfigure	Prefatigue	Postfatigue
20 - 25				
50 - 60				

After collection of all the data, a repeated measures analysis of variance (ANOVA) was conducted on each set of 3 impact trials that made up each of the 8 cells in the above table, to evaluate any significant differences between the trials ($\alpha \leq 0.05$). Post-hoc analysis was done for all significant effects using paired t-tests with a Bonferroni adjustment. A three factor analysis of variance (ANOVA) was then completed on each dependant variable, to identify significant interactions and main effects, using an alpha level of $p \leq 0.05$ for all comparisons.

Based on the statistical procedures used, the following null hypotheses are warranted:

Null Hypothesis # 1: There will be no significant differences between pre and post-fatigue measurements of peak acceleration, acceleration slope or acceleration time to peak when the localized shank muscles are fatigued.

Null Hypothesis # 2: There will be no significant difference between post-fatigue measurements of peak acceleration, acceleration slope or acceleration time to peak when fatigue of the dorsiflexors is compared to fatigue of the plantarflexors.

Null Hypothesis # 3: Post-fatigue measurements of peak acceleration, acceleration slope or acceleration time to peak will show no significant differences when women in the older (50 – 60 yrs) age group are compared to women in the younger (20 – 25 yrs) age group.

Chapter IV

RESULTS

4.1 Trial Repeatability

There was no main effect for trial for all dependant variables analyzed ($p = 0.054$ to 0.838). Impact velocity differences between trials were also found to be not statistically significant, except for the 3 post-fatigue trials in Session 2. These showed a significant difference between trials 1 and 2 ($p = 0.008$); and between trials 1 and 3 ($p = 0.0002$). However, all of the velocities and/or impact forces fell within the target regions of $1 - 1.15$ m/s and $1.8 - 2.8$ X BW, respectively. Since these were the only significant differences found between trials, mean values of the three trials were used for further analysis of all between subject and within subject variables.

There were also no significant differences for all dependant variables (PA, TP and AS) for the pre-fatigue trials between Session 1 and Session 2 ($p = 0.392$ to 0.841).

4.2 Dependant Variables

4.2.1 Peak acceleration (PA)

The magnitudes of the mean peak accelerations are outlined in Table 3. As can be seen in Figure 11, the mean peak acceleration measured at the tibial tubercle decreased significantly after fatigue ($p = 0.008$). Although not statistically significant, there was a

Table 3. Mean peak acceleration \pm S.D. (g's)

Age group (years)	Pre-fatigue Session 1	Post-fatigue Session 1	Pre-fatigue, Session 2	Post-fatigue Session 2
20 - 25	12.96 ± 4.7	11.27 ± 3.5	11.86 ± 3.4	11.06 ± 2.9
50 - 60	13.58 ± 2.8	12.77 ± 3.8	14.69 ± 5.2	12.93 ± 3.9
total	13.28 ± 3.7	12.09 ± 3.1	13.21 ± 4.5	11.95 ± 3.5

trend towards higher levels of acceleration being measured in the older age group. There were no other significant main effects or interactions for this variable.

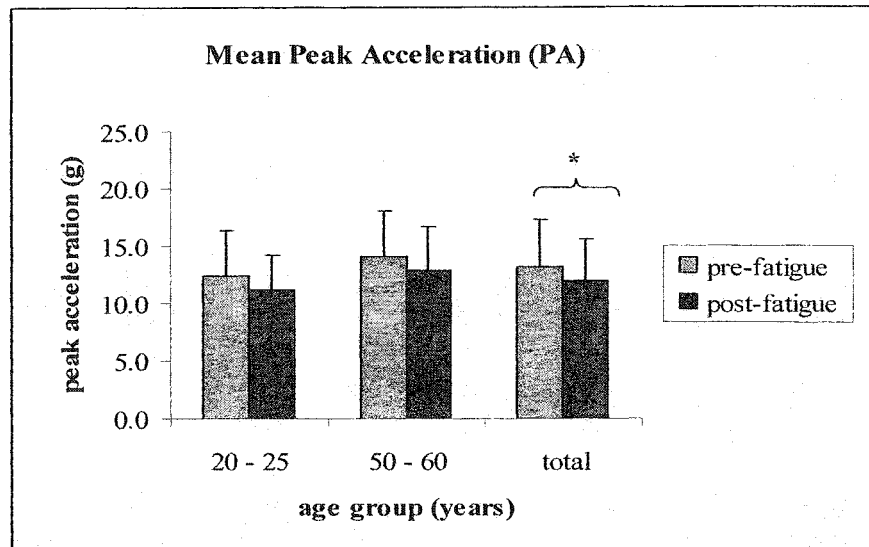


Figure 11. Mean Peak Acceleration (PA), pre vs. post-fatigue. (* $p = 0.008$)

4.2.2 Acceleration slope (AS)

Mean acceleration slopes are outlined in Table 4. Similar to peak acceleration, the average slope was significantly lower after fatigue ($p = 0.033$), as demonstrated in Figure 12. Again, the older women demonstrated a higher, though not statistically significant, acceleration slope. There is a high standard deviation in acceleration slope indicating the significant variation found between subjects.

Table 4. Mean acceleration slope \pm S.D. (g/s)

Age group (years)	Pre-fatigue Session 1	Post-fatigue Session 1	Pre-fatigue, Session 2	Post-fatigue, Session 2
20 - 25	2830 \pm 1972	1963 \pm 1515	2322 \pm 1662	2020 \pm 1444
50 - 60	3265 \pm 976	2794 \pm 1152	3416 \pm 2029	3138 \pm 1979
total	3067 \pm 1488	2416 \pm 1363	2843 \pm 1883	2589 \pm 1759

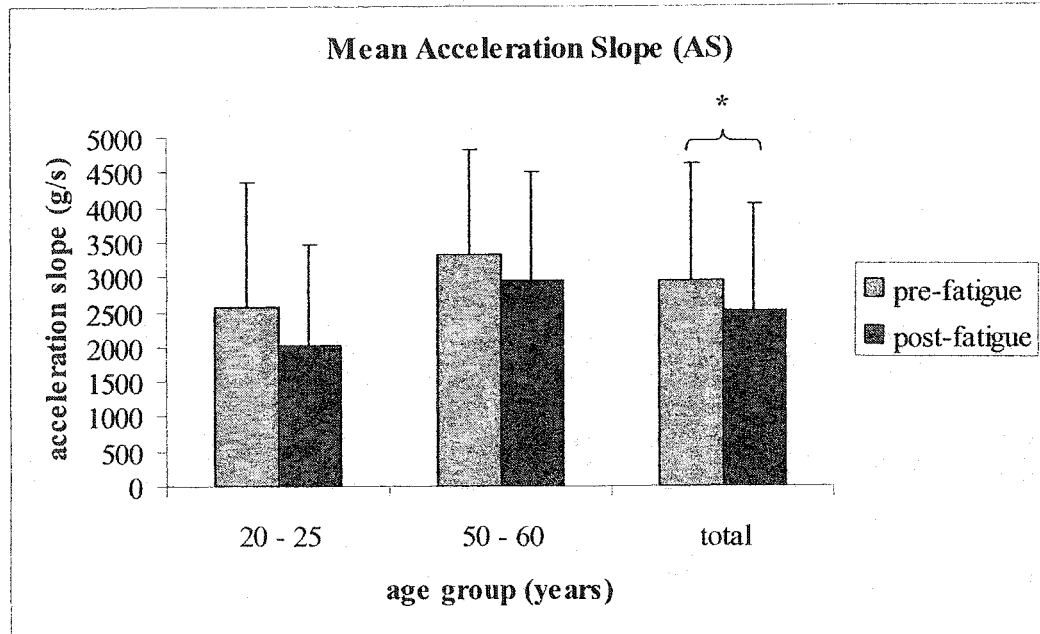


Figure 12. Mean Acceleration Slope (AS), pre vs. post-fatigue (* $p = 0.033$)

4.2.3 Acceleration time to peak (TP)

Average times to peak acceleration for the different sessions and age groups are outlined in Table 5. There were no significant main effects or interactions for this variable.

Table 5. Mean acceleration time to peak \pm S.D. (s)

Age group (years)	Pre-fatigue Session 1	Post-fatigue Session 1	Pre-fatigue Session 2	Post-fatigue Session 2
20 - 25	0.0102 ± 0.003	0.0118 ± 0.005	0.0107 ± 0.003	0.0126 ± 0.004
50 - 60	0.0101 ± 0.007	0.0106 ± 0.006	0.0086 ± 0.002	0.0076 ± 0.002
total	0.0101 ± 0.005	0.0109 ± 0.006	0.0097 ± 0.002	0.0102 ± 0.004

4.3 Impact Parameters

4.3.1 Impact force

No significant main effects or interactions were found for impact force peak. Table 6 outlines the mean impact force peaks for the two age groups and varying input parameters.

Table 6. Mean peak impact force \pm S.D. (X BW)

Age group (years)	Pre-fatigue Session 1	Post-fatigue Session 1	Pre-fatigue Session 2	Post-fatigue Session 2
20 - 25	2.43 ± 0.33	2.34 ± 0.30	2.49 ± 0.31	2.42 ± 0.33
50 - 60	2.36 ± 0.40	2.30 ± 0.40	2.31 ± 0.33	2.19 ± 0.24
total	2.39 ± 0.11	2.32 ± 0.35	2.40 ± 0.33	2.31 ± 0.31

As shown in Figures 13 and 14, the correlations between pre-and post-fatigue impact forces in Session 1 and Session 2 were fairly high, with R^2 values of 0.501 and 0.714, respectively. This indicates acceptable consistency of impact force averages within each impacting session.

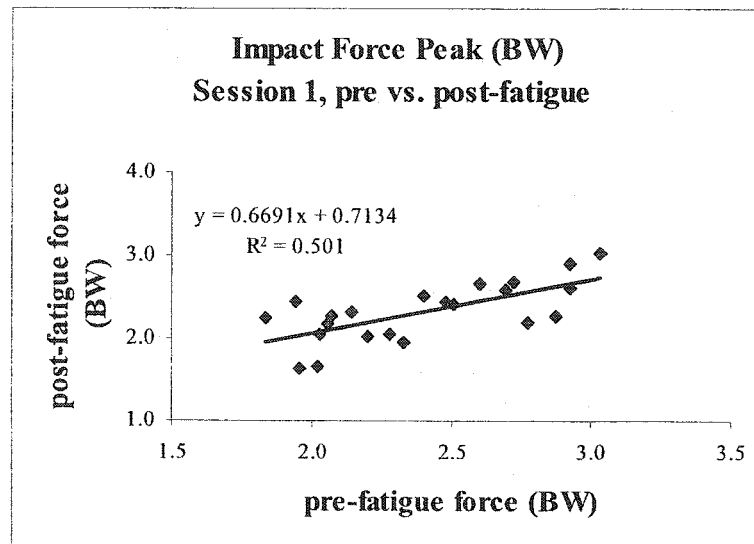


Figure 13. Impact force peak correlation, Session 1, pre vs. post-fatigue

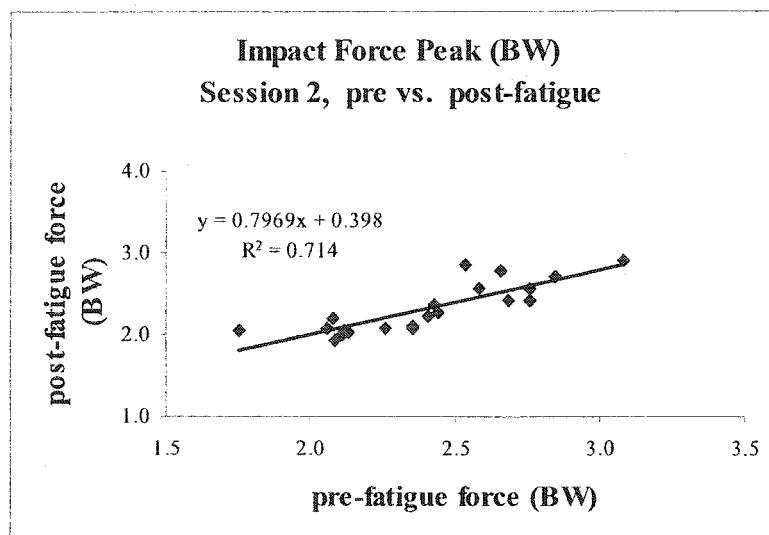


Figure 14. Impact force peak correlation, Session 2, pre vs. post-fatigue

4.3.2 Impact force/velocity/acceleration

Figure 15 illustrates the significant variability in peak tibial acceleration between subjects. This variability occurred despite fairly consistent input parameters of velocity and peak force. As mentioned earlier, some of the data from the first session had to be removed due to the subjects' inability to maintain dorsiflexion during the post-fatigue

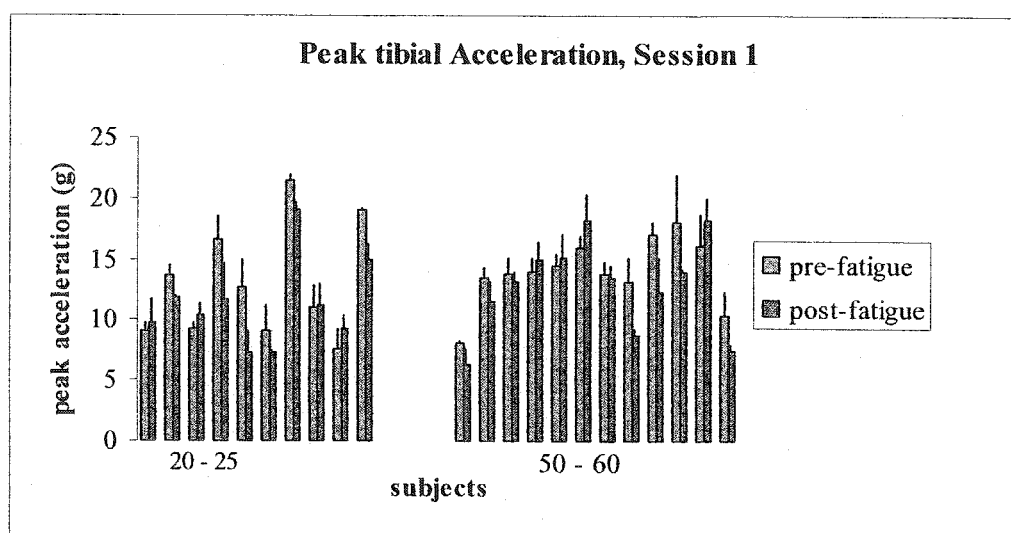


Figure 15. Peak tibial acceleration \pm S.D. for each subject, pre and post-fatigue during Session 1

impacts. Thus, there appear only 10 subjects in the younger age group for Session 1.

Figure 16 shows the average velocities of the pendulum at the time of impact, for each subject, in Session 1. All but three of the velocity measurements fell within the 1 – 1.15m/s target velocity. Peak impact forces for the same impacts are shown in Figure 17. Of note are two instances where the impact force fell below the target range post-fatigue yet the velocity changed minimally or increased (see symbols † and ‡, respectively).

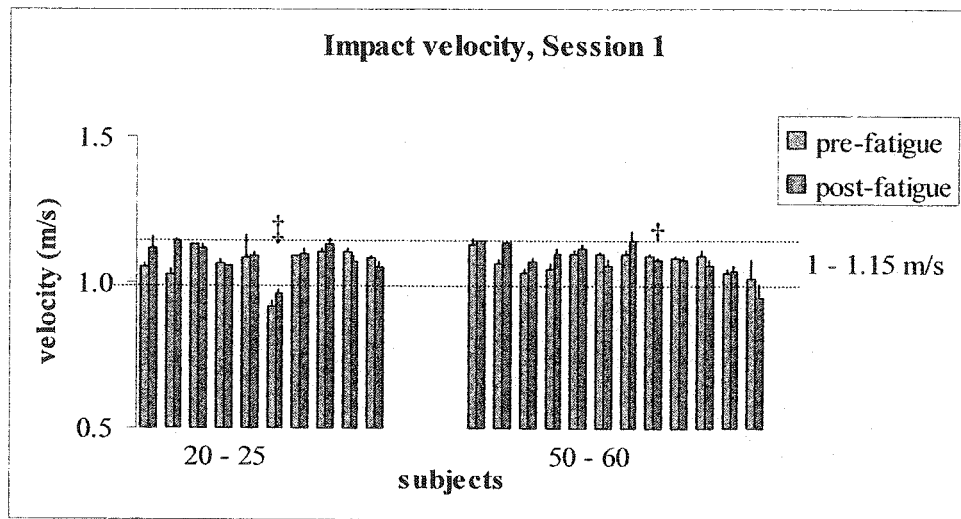


Figure 16. Mean impact velocities \pm S.D. for each subject, pre and post-fatigue during Session 1.

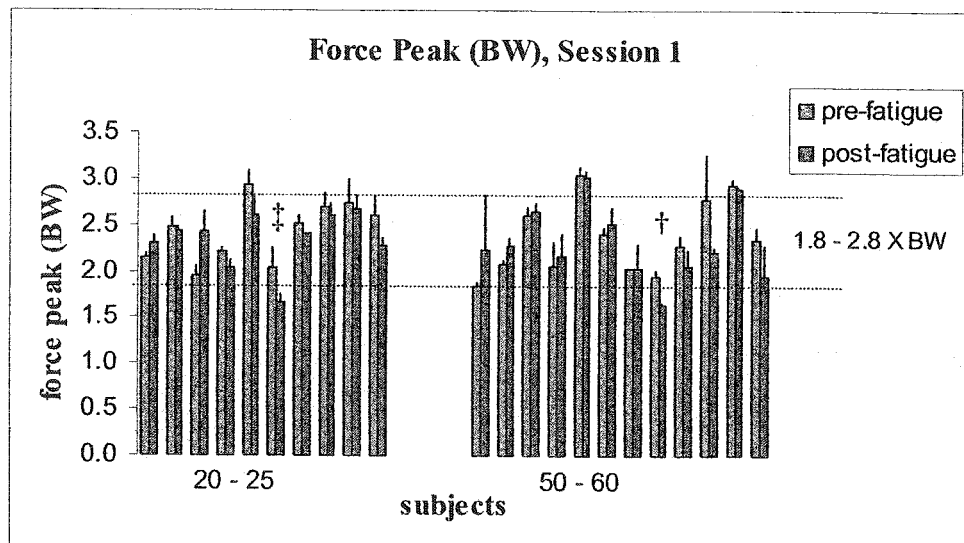


Figure 17. Mean impact force peaks \pm S.D. for each subject, pre and post-fatigue, during Session 1.

4.3.3 % MVC of tibialis anterior at impact

As shown in Figure 18, there was a significant increase in the % MVC of tibialis anterior after fatiguing exercise ($p = 0.003$). In session 2, when gastrocnemius was fatigued, the tibialis anterior % MVC decreased minimally, indicating a significant main effect for Fatigue ($p = 0.003$) as well as a Fatigue*Session interaction ($p = 0.016$).

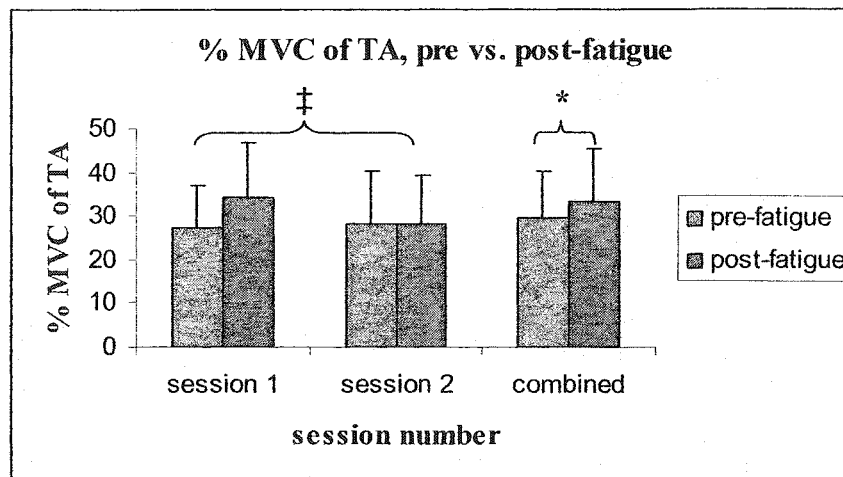


Figure 18. % MVC of Tibialis Anterior (TA) \pm S.D., pre vs. post-fatigue; Session 1 and Session 2. (* $p = 0.003$, ‡ $p = 0.016$ interaction)

4.3.4 %MVC of gastrocnemius at impact

There were no significant main effects or interactions for gastrocnemius % MVC at impact. However, the % MVC of gastrocnemius on both days did tend to increase after fatigue, as shown in Figure 19.

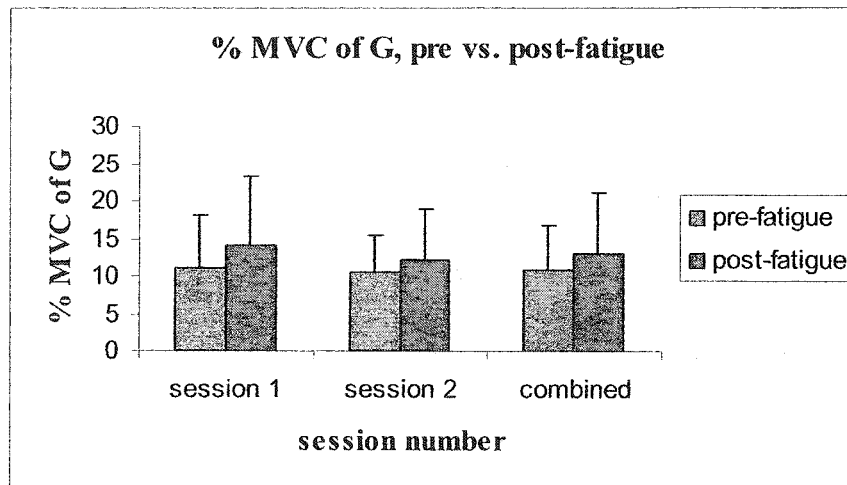


Figure 19. % MVC of Gastrocnemius (G) \pm S.D., pre vs. post-fatigue; Session 1 and Session 2.

4.3.5 Fatigue

Underlying this experiment is the assumption that the tibialis anterior and gastrocnemius muscles were actually being fatigued during the isometric exercise. A decrease of at least 15% in the MPF was being used as a benchmark, to ensure that the muscles were in a fatigued state (as per the work of Ament et al., 1993).

Table 7 outlines the changes in MPF found in each session. The two fatiguing bouts completed during each session are indicated by Fatigue 1 and Fatigue 2 in Table 7. The older age group did not demonstrate as large a % change in MPF as the younger group, and on average took approximately 20% longer to fatigue.

Table 7. Mean fatigue-induced changes in MPF \pm S.D. Changes are calculated relative to initial MPF prior to the first bout of fatiguing exercise. Results are listed as absolute change (Hz), as well as % change.

Age group	Session 1 (fatigue TA)				Session 2 (fatigue G)			
	Fatigue 1		Fatigue 2		Fatigue 1		Fatigue 2	
	Hz	%	Hz	%	Hz	%	Hz	%
20 - 25	-32 \pm 14	-24 \pm 9	-44 \pm 20	-32 \pm 12	-32 \pm 18	-21 \pm 10	-45 \pm 20	-30 \pm 11
50 - 60	-33 \pm 10	-23 \pm 6	-33 \pm 13	-24 \pm 9	-20 \pm 11	-13 \pm 7	-23 \pm 19	-14 \pm 13

Chapter V

DISCUSSION

5.1 Overview

The first purpose of this research was to analyze the effect of localized muscle fatigue on the leg's ability to attenuate impact shock waves as they travel from the heel to the knee. From this study, it appears that local fatigue causes an increase in shock attenuation, as seen by a decrease in the peak acceleration being measured at the tibial tubercle, as well as a decrease in the acceleration slope. This is in direct contrast to many studies that measured acceleration before and after full body or cardiovascular fatigue, and found a decrease in attenuation ability (Milgrom, 1989; Mizrahi et al., 1997; Fyhrie et al., 1998; Verbitsky et al., 1998; Voloshin et al., 1998; Mizrahi et al., 2000a; Mizrahi et al., 2000b; Mizrahi et al., 2001). Therefore, it appears that full body fatigue does not affect attenuation properties in the same way as local muscle fatigue.

The second purpose of this study was to analyze the effect of aging on shock attenuation. Statistical analysis showed that there was no significant effect for age on any of the dependant variables. However, there are some interesting trends, and these will be discussed.

5.2 Hypotheses Revisited

#1. Null Hypothesis: *Localized fatigue of the plantarflexors or dorsiflexors will have no effect on the attenuation properties of the shank.*

The null hypothesis is rejected.

***Research Hypothesis:** As the muscles of the leg fatigue, they will be less able to attenuate shock. Therefore, the magnitude of the peak acceleration and the rate of the increase of acceleration being measured at the tibial tubercle at the time of impact will increase. The time taken to reach peak acceleration will decrease.*

Based on the findings of this research, the research hypothesis cannot be accepted.

Previous research has found that full body fatigue causes a decrease in attenuation properties of the shank. It, therefore, seemed reasonable to expect that localized muscle fatigue would result in the same changes. Voloshin et al. (1998) noted an increase in tibial acceleration of approximately 60% after 30 minutes of running, and attributed this change to the inability of the fatigued system to efficiently absorb shock waves.

Verbitsky et al. (1998), after having subjects run on a treadmill to induce fatigue (based on PETCO₂ levels), also noted an increase in acceleration amplitude. He added, "...the human musculo-skeletal system becomes less capable of handling heel strike-induced shock waves when the muscles are significantly fatigued." The general premise of these articles is that fatigue causes an increase in shock transmission.

Mizrahi et al. (1997) recognized that full body fatigue, as indicated by increasing PETCO₂ levels, is not reflected by changes in shank muscle EMG levels. He suggested that the decrease in shock attenuation was due to a deterioration of coordinated muscle activity and loss of synergistic muscle control more so than individual muscle fatigue. The coordinated exchange of energy between eccentric and concentric phases of running provides an efficient shock-absorbing mechanism (Mizrahi and Susak, 1982; Derrick et al., 1998). General fatigue interferes with this exchange, causing a reduction in the angular displacement of the weight-bearing joints with a corresponding increase in the

angular velocity during the eccentric phase. The concentric phase tends to decrease in angular velocity. The overall effect of these changes is a gradual increase in stiffness of the limb (Gollhofer et al., 1987), which causes increased shock propagation.

In contrast, the human pendulum technique allows for control of joint angles, thus eliminating most of the relevant angular velocities. The control of foot dorsiflexion during impact still requires eccentric muscle activity, and the quadriceps are required to maintain knee extension. However, these are both physiological requirements of running, and were not seen as detracting from the relevance of the study. There were also no significant effects for Session in any of the dependant variables, thus control of foot dorsiflexion after fatigue of tibialis anterior was not found to be a significant problem.

Using the pendulum method, it appears that as the local muscles of the leg fatigue, they tend to attenuate significantly more shock, causing a decrease in the magnitude of the tibial peak acceleration as well as the acceleration slope. The time taken to reach peak acceleration did not differ significantly with fatigue, but it was seen to increase in both age groups when tibialis anterior was fatigued, and in the younger women when gastrocnemius was fatigued.

Local muscle fatigue is characterized by a decrease in force-generating capacity (Kent-braun, Ng, Doyle and Towse, 2002) and a decrease in tension (Winter, 1991). These changes cause the muscle to be less stiff, thus more likely to attenuate force. In the same way that the mechanical model by Pain and Challis (1991) demonstrated that a rigid structure transmitted more acceleration than one with a wobbling mass component, it can be extrapolated that as the wobbling mass becomes less stiff, it will absorb even

more of the shock. This will lead to a decrease in the peak acceleration reaching the tibial tubercle, and a concurrent decrease in the acceleration slope. However, as was found in this study, a significant change in the time the shock wave takes to travel up the leg is not evident.

2. Null Hypothesis: *There will be no difference in shock attenuation properties when localized fatigue of the dorsiflexors are compared to fatigue of the plantarflexors.*

This hypothesis cannot be rejected.

Knowing that the muscles of the leg make up the major portion of the wobbling mass, it seemed reasonable to assume that alterations in the muscle tension would be manifested by changes in attenuation ability. The plantarflexors are a much larger muscle group than the dorsiflexors, thus any main effect demonstrated with fatigue should be greater with plantarflexor fatigue.

In fact, the differences between Sessions in force attenuation changes were not statistically significant. Fatigue of the tibialis anterior caused a decrease in peak tibial acceleration of 1.19 g's (9.03%), while fatigue of the gastrocnemius caused a similar decrease of 1.26 g's (9.54%). Thus, the trend was for plantarflexor fatigue to have a slightly larger effect than dorsiflexor fatigue, which does lend support to the hypothesis.

The slope of the acceleration curve, although not significant, did show a tendency to have the greatest change post-fatigue when the dorsiflexors were fatigued, as opposed to the plantarflexors. This was probably due to the nature of the study design. The subjects were expected to maintain ankle dorsiflexion prior to and during impact, to prevent the entire foot from slapping against the force plate. Fatigue of the tibialis anterior muscle may have interfered with the subjects' ability to maintain dorsiflexion,

resulting in more of the heel pad contacting the force plate at impact. This extra damping would have resulted in a lower acceleration slope. Further research might consider supporting the forefoot, thus eliminating the requirement of foot dorsiflexor activity during impact. It was decided not to include such an apparatus in this study, as it would have been difficult to support the forefoot yet allow fatiguing exercises, rapidity of testing post-fatigue, free movement of the soft tissues of the shank, and not interfere with the data collection.

The average time to peak acceleration when the two age groups were combined tended to increase on both days. However, the individual age groups responded differently with this variable. The younger women demonstrated a greater increase in the time to peak on the second day; thus, when gastrocnemius was fatigued, the impact shock took longer to reach the tibial tubercle. This supports the hypothesis of a larger muscle fatigue effect with plantarflexion fatigue. The older group showed a slowing down of shock transmission when the dorsiflexors were fatigued. However, a faster time to peak was evident when the plantarflexors were fatigued, possibly due to the heel pad effect mentioned above.

3. Null Hypothesis: There will be no difference in attenuation properties in the shank between the younger (20 – 25 years) and older (50 – 60 years) groups of women.

This hypothesis cannot be rejected. The statistical analysis revealed no significant effect for age on any of the dependant variables. However, once again, there are trends.

In Session 1, when the dorsiflexors were fatigued, the younger group demonstrated greater differences in all three dependant variables than the older group. The younger women also underwent a larger shift in the MPF of the EMG towards lower

frequencies, indicating greater levels of fatigue. On both days, the older women consistently took longer to fatigue than the younger group. Previous studies have found greater fatigue resistance in older subjects (Chan, Raja, Strohschein and Lechelt, 2000; Kent-Braun et al., 2001). This is thought to be due to a gradual increase in the % of Type 1 (slow-twitch) muscle fibres compared to Type 2 (fast-twitch) with aging (Lexell, Henriksson-Larsen, Winblad, Sjöström, 1983). Lexell et al (1983) found an almost 50% atrophy in the cross-sectional area of the Type 2 fibres, as well as an actual decrease in their number. Combining these findings, they calculated that the Type 2 fibres occupied 55% of the relative area in a young muscle, but only 30% of the area in an aged muscle. The older women, then, have a larger % of the muscle area made up of Type 1 fibres, and will be less likely to manifest dramatic changes in muscle tension as they fatigue.

Considering these effects of aging and gastrocnemius's greater fatigue resistance (taking almost twice as long to fatigue as TA, on average), one would expect that the young women on Session 1 would manifest the largest pre and post-fatigue differences in the 3 dependant variables. In fact, the young women did consistently show the largest differences if Session 1 is considered alone. However, in Session 2, when the gastrocnemius was fatigued, there was no consistent pattern.

5.3 Comparing with Previous Studies

Tibial impact acceleration has been measured many different ways in previous studies. The two main differences in study design involve: 1) whether the subjects are shod or barefoot, and 2) having the subjects physically running during data collection. We chose to impact the bare foot of the subjects, to eliminate the variability that different shoe types would bring to the study. However, we did one additional impact, at

the end of each session, with a shoe on their foot. We used the subject's own shoe that they had worn to the session, or a running shoe if they had brought one.

The average change in the impact force, as shown in Table 8, was approximately 0.5 X BW. Our higher impact forces may also result in acceleration peaks that are higher than those found in shod experiments, with associated lower times to peak and a greater acceleration slope.

Table 8: Comparisons of mean impact forces, barefoot vs. shod

Age group	Impact Force (BW) barefoot	Impact Force (BW) Shod	Difference in Impact Force (BW)
20 – 25:	2.43	1.90	0.53
50 – 60:	2.34	1.81	0.53
Total:	2.38	1.85	0.53

Table 9 compares our mean pre-fatigue results with those of previous studies. Kim et al (1994) found acceleration peaks of 60 g's at the heel, which were reduced to approximately 12 g's at the tibial tubercle in a running barefoot subject. Time to peak and rate of loading were not mentioned in their study. Lafortune et al. (1996) describe a mean peak acceleration of 12.5 g's with barefoot pendulum impacts, which increased to 19 g's when the knee was bent to 40° of flexion. Dickinson et al. (1996) observed ground reaction forces in shod and barefoot runners. They found that, with shod running, the impact force peak occurred within 23 ms of impact, but this dropped to approximately 10 ms when the subject ran barefoot. Assumedly, the time taken to reach peak acceleration would also have dropped, which would explain our low time to peak results.

Table 9. Comparisons of mean outcomes with previous studies. PF = peak impact force, PA = peak acceleration, TP = time to peak acceleration, AS = acceleration slope

	PF (bw)	PA (g)	TP (ms)	AS (g/s)
20 – 25 (current study)	2.4 ± 0.3	12.4 ± 4.0	10.45 ± 5	2576 ± 1817
50 – 60 (current study)	2.3 ± 0.3	14.1 ± 4.0	9.35 ± 3	3340 ± 1502
Lafortune & Lake, 1995; shod, pendulum	2.0 ± 0.2	6.4 ± 0.7	16.1 ± 3	671 ± 220
Kim et al., 1994 bare-foot, running	-----	$\cong 12$ 60 at the heel	-----	-----
Lafortune et al., 1995 Shod, running	-----	11.2 ± 3.1	31 ± 6	-----
Lafortune et al., 1996 a) Shod, pendulum b) barefoot, pendulum	-----	9.4 ± 4 $\cong 12.5$	22.5 ± 5.5 -----	1150 ± 930 -----
Verbitsky et al., 1998 Shod, running	-----	10 ± 1.25	-----	-----

Thus, the outcome measures from this study are in keeping with the current literature, given that there is little written about impact acceleration with barefoot running. The older women tend to have a slightly shorter time to peak accompanied with higher rate of loading and peak acceleration. This may be partly explained by the decreasing elasticity found in heel pads as people age (Hsu, Wang, Tsai, Kuo and Tang, 1998). Decreasing muscle bulk and bone density may also be playing a role in these outcomes.

5.4 Variability

As shown by the large standard deviations associated with the three dependant variables, there was large between subject variability. However, similar to the work done by Andrews and Dowling (2000), there were also many instances of within subject variability. For example, an exact impact velocity repeated consecutively on the same

subject resulted in impact forces which changed as much as $0.22 \times BW$. In numerous instances, a decrease in impact velocity caused an increase in force peak. Minimal changes in peak impact force could cause a moderate change in peak tibial acceleration. For example, with one subject, a 5% decrease in impact force led to a 21% reduction in peak tibial acceleration.

There appear to be many other variables that affect shock attenuation in the shank. As mentioned previously, the exact position of the heel pad relative to the force plate at impact alters the total surface area that is available to dissipate the impact force. The % MVC of both muscle groups will potentially affect the total segment stiffness. In this study, there were instances (within a set of 3 consecutive impacts) of % MVC being proportional to PA, yet there were also instances where an increased % MVC was observed when the PA decreased. The concept of muscle “tuning”, as described by Wakeling and Nigg (2001a) suggests that the level of tension in a muscle is based on previous experiences of input frequency, amplitude and time, in an attempt to minimize soft tissue resonance. The perception or anticipation of discomfort will, therefore, affect muscle tuning, and is extremely difficult to control when using human subjects.

5.5 Physiological Perspective

EMG analysis of muscle fatigue demonstrates a decrease in the mean power frequency (MPF), as well as an increase in the amplitude of the signal (Winter, 1991). In this study, the % MVC (as measured by EMG signal amplitude) increased with fatigue in most cases, most notably when the tibialis anterior was fatigued. Consequently, one might be led to think that increasing % MVC would indicate more tension and stiffness

in the muscle. However, fatigue is also manifested by a decrease in muscle tension, given that the muscle activation remains the same (Winter, 1991).

The significant findings in this study of lower peak acceleration and slope of acceleration peak, as well as the tendency of an increased time to peak after fatigue all indicate that the muscle becomes less stiff. A less stiff segment will attenuate more of the impact acceleration than one with increased stiffness. How, then, do we explain this paradox of increasing signal amplitude with decreasing muscle stiffness?

According to Winter (1991), fatigue is known to alter the shape of the motor unit action potential (m.u.a.p.) in the following ways:

- Lower conduction velocity
- Slower rise and fall of the of the action potential spike
- A gradual shift to the slower m.u.a.p.'s remaining active, while the faster motor units with shorter m.u.a.p.'s gradually drop out
- Increased synchronization of the motor unit firing

Consequently, the EMG spectrum will shift to those of lower frequencies, but with higher amplitude.

Petrofsky et al. (1982) demonstrated that the MPF of a muscle is independent of the force being generated. Thus, higher amplitude in the EMG spectrum does not necessarily indicate greater force, especially during fatiguing exercise. DeLuca (1979) found that fatigue caused the EMG amplitude from surface electrodes to increase, while the amplitude from indwelling electrodes decreased. The EMG signal is a summation of motor unit action potentials from many myofibrils beneath the skin. They may not all react the same way.

Fitts (1995) described a possible explanation for this decrease in force associated with higher EMG spectrum amplitude. With fatigue, the resting potential of the surface membrane gradually depolarizes from the normal resting potential of -80 mV to about -70 mV. The action potential spike height decreases by approximately the same amount, and increases in duration. However, deep in the t-tubule, there may well be a greater depolarization (to approximately -60 mV), and the action potential spike lowers to the extent that it fails to reach the threshold of 0 mV. Consequently, only the superficial regions of the muscle fibre react to the activation signal from the motoneuron. The deeper regions do not have a high enough action potential relative to the threshold required to elicit the release of calcium and subsequent actin/myosin interaction, which forms the basis of muscle force generation. The end result would be a high surface EMG signal amplitude relative to the low muscle contraction force deep within the muscle fibre.

5.6 Functional Implications of the Research Findings

The wobbling mass component of the body is known to absorb much of the shock wave that travels up the body after heel-strike. This research illustrates that, as the specific muscles of the shank fatigue, they are better able to attenuate shock. Contrary to the effects of central, or cardio-vascular fatigue, which cause a decrease in attenuation abilities, localized muscle fatigue will tend to protect the joints which lie proximal to the shank, by absorbing more of the impact. This would become particularly important when running for prolonged periods of time on a decline or incline, causing fatigue of the dorsiflexors, or plantarflexors, respectively. The significance of this work is also important to people who are required to spend a lot of time on their toes or heel, such as

dancers, especially when forceful landing impacts are a frequent occurrence in their routine. These findings may also be of interest to researchers investigating fatigue in other regions of the body, such as the hand and arm, when impacts are a required activity.

5.7 Limitations and Assumptions

The human pendulum method was used in an attempt to control joint angles and velocities, allowing a more specific look at the effect of localized muscle fatigue. Lafortune and Lake (1995) demonstrated that the pendulum simulates initial impact conditions similar to those found in running. However, the ankle joint angle was difficult to control, especially post-fatigue during Session 1. Impacts with a flat foot, instead of the heel, caused lower impact forces, as well as lower peak acceleration in numerous trials, and were recognized in the data by a lower or absent initial impact peak on the force/time graph. The majority of subjects were able to produce at least 3 trials with a clear heel strike pattern. The data from 2 subjects from Session 1 had to be discarded, due to poor dorsiflexion control (and thus, poor heel strike). However, there were instances where the subjects were able to maintain an adequate heel strike, but not to the same degree as pre-fatigue. This may have affected the results slightly, but, as mentioned before, there was no significant effect for either Session in this study. Future studies may choose to control ankle joint angle, or measure it at the time of impact, incorporating the changing angle as another variable.

In normal running, however, we do not absorb all of the impact force with our heel; the forefoot contacts the ground and propels us forward. The relevance of this

experiment is based on the assumption that results obtained from a heel-only impact will also pertain to a normal heel-toe gait.

Despite using an upright metal barrier as a landmark, it was difficult to pull the pendulum back the exact same distance with each trial. A magnetic release system, firmly attached to the wall or floor, might have decreased the variability between trials, by consistently reproducing the same distance.

On average, the first post-fatigue impact occurred 20 seconds after the fatigue apparatus was removed. This was longer than proposed, but was found necessary due to the specifics of our software system. Based on the work of Rohmert (1973), muscles held at 50% of their MVC for 66 seconds (our average time to fatigue in Session 1) require over 11 minutes to recover. In Session 2, the average time to fatigue was 116 seconds, which would have required far more recovery time. The assumption was therefore made that the muscles were still fatigued at the time of all impacts.

The “older” group in our study were recruited from women with a maximum of 60 years of age. However, the oldest woman was 57, and the average age was 53.4. Even at this age, it was difficult to find subjects who did not have medical or joint complaints which precluded their participating in the study. Ideally, when comparing them to young adults, it would have been ideal to have an older group in their 70’s. However, the impacting nature of this study made that unreasonable, if not dangerous. Therefore, we were limited in our ability to compare subjects from two vastly different age groups, and this may have diluted any possible main effect for age.

We also assumed that, on average, the older women were of a similar fitness level as the younger ones. The average BMI (mass/height²) was 24.4 ± 2.4 for the older

women and 22.2 ± 3.1 for the younger women. Both fall within the “Canadian Guidelines for Body Weight Classification in Adults” (2003), published by Health Canada, which recommends a BMI between 18.5 and 24.9. Four of the older women and two of the younger ones fell slightly above the recommended BMI, yet all but one admitted to regular exercise routines which involved activities such as weight training, running, volleyball or curling. The guidelines from Health Canada also note that persons 65 and over, with a slightly elevated BMI, may still be considered within the ideal range.

5.8 Future Directions

Peak tibial acceleration following impact of the heel decreased as the local muscles of the shank fatigued. There was no significant difference when dorsiflexor fatigue was compared to plantarflexor fatigue. In the future, the effect of fatiguing both muscle groups simultaneously, in a ratio representative of that found in running, may demonstrate a change which better represents that found in normal running.

Although there was no significant main effect for age, the older women did tend to have higher peak acceleration and acceleration slope than the younger group. Future research might focus on an older group of women (ie 70 – 75 years), yet lower the impact force and velocity to a range which is easily tolerated and safe. Providing identical shoes to these women and allowing some knee flexion would decrease total force transmitted, better simulate running, while standardizing the shock-absorbing qualities of the shoe.

Other variables that would be interesting to compare in the future include:

- BMI: compare those with high BMI to those whose BMI falls within either the ideal, or the low category

- Activity level: compare sedentary women to a group who participates regularly in weight-bearing exercises.
- Shank height: establish what relationship exists between tibial length and attenuation of acceleration.
- Bone density: using bone densitometry as our indicator, determine the effects of decreasing bone density on shock absorption.
- % MVC of the shank muscles: look at how varying muscle tension affects shock attenuation.

5.9 Conclusions

In this experiment, the effect of localized fatigue of the dorsiflexors and plantarflexors on tibial impact acceleration was analyzed. Using PETCO₂ as an indicator, many previous studies have found that full body fatigue caused a decrease in shock attenuation properties, thus acceleration measured at the tibial tubercle increased with fatigue (Milgrom, 1989; Mizrahi et al., 1997; Verbitsky and Isakov, 1998; Fyhrie et al., 1998; Voloshin et al., 1998).

However, with localized muscle fatigue, there appears the opposite effect: fatigue of the plantarflexors or dorsiflexors causes a significant increase in attenuation properties of the shank. The magnitude of peak acceleration and acceleration slope measured at the tibial tubercle demonstrated a significant decrease with localized muscle fatigue. From this research, it appears that full body fatigue cannot be equated with localized muscle fatigue when analyzing shock attenuating properties of the leg.

In addition, this research did not demonstrate any statistically significant difference in shock attenuating properties between younger (20 – 25 years) and older (50 – 60

years) women. Nor did fatiguing the dorsiflexors demonstrate any significantly different effect than fatigue of the plantarflexors.

Previous literature states that, as a muscle fatigues, it has a decreased capacity for force generation, and is less able to generate muscle tension (Winter, 1991). Based on the findings of this experiment, it appears that localized muscle fatigue of the plantarflexors or dorsiflexors results in an increase in the shock attenuating properties of the shank.

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Appendix 1: Poster for recruitment of subjects

VOLUNTEERS REQUIRED FOR RESEARCH AT UNIVERSITY OF WINDSOR

I am asking for female volunteers to participate in a study for my Masters' thesis entitled, "*The effect of localized muscle fatigue on tibial shock acceleration, using the human pendulum method*".

In normal running, the impact force of the heel hitting the ground is transmitted through the body. Numerous factors such as knee angle, velocity and fatigue can affect the amount of the impact which the body absorbs, in its attempt to soften the impact effect on the back, neck and head. Although a lot of research has been done on fatigue and shock absorption, it has all been done with the subjects actually running, which does not control other factors such as joint angles, muscle forces or speed. I plan to have subjects lie down on a bed that is suspended from the ceiling, with their foot extending over the end of the bed. The bed will be pulled back and released, allowing them to swing forwards until their foot impacts against a vertical force plate. This will be repeated after the anterior shin muscles (the dorsi-flexors) are fatigued, to see the effect fatigue has on force absorption. In a second test session, the same subjects will repeat the above impact testing, but fatiguing the calf muscles (plantar-flexors) instead.

Women between the ages of 20 and 25 will be compared to women between 50 and 60, to see if this effect changes with aging.

WHO CAN VOLUNTEER?

I am looking for females between the ages of 50 and 60 who participate in some form of weight-bearing exercise (ie running, aerobics, step-class, elliptical trainer) at least twice a week. Women with recent or chronic joint pain in the leg, back or neck are not suitable due to potential aggravation of their pain.

HOW LONG WILL IT TAKE?

Each participant will be required to attend for two sessions, each lasting about 1 hour. The study will hopefully begin in March, 2003. Hours are very flexible.

WILL IT HURT?

Although the forces used will be similar to those measured in normal running, the hip and knee will be kept straight, thus the force will feel greater. However, it is still within a range that is safe.

DO I GET PAID?

No, this is entirely a volunteer project.

WHO DO I CONTACT?

Janice Flynn, registered physiotherapist
Masters' candidate
Faculty of Human Kinetics, U of Windsor
253-3000 ext.2468 (university)
258-7836 (home)
dandrews@uwindsor.ca
email: flynn3@uwindsor.ca

Dr. David Andrews,
Associate Professor
University of Windsor
253-3000 ext. 2433
email:

Appendix 2: Subjects' Consent to Participate in Research Form

CONSENT TO PARTICIPATE IN RESEARCH

The Effect of Localized Muscle Fatigue on Impact Shock Attenuation in the Leg, Using the Human Pendulum Method of Impact

You have been asked to participate in a research study conducted by: Dr. David Andrews, Janice Flynn and Jeff Holmes at the University of Windsor.

If you have any questions or concerns about the research, please feel free to contact Dr. David Andrews, Associate Professor, Faculty of Human Kinetics, University of Windsor (253-3000, X 2433; Room HK Building; dandrews@uwindsor.ca).

PURPOSE OF STUDY

The purpose of the study is to determine what effect fatigue of the lower leg muscles has on the amount of shock transmitted from the foot through to the knee region. The amount of impact shock will simulate that found in running. Women of two differing age groups will be compared, to analyze if a fatiguing effect alters with aging.

PROCEDURES

The study will occur over two days, which are several days apart.

On day 1, a maximal contraction of the anterior shin muscles will be elicited, while being analyzed by electromyography (EMG). To ensure a good signal, the skin will be shaved and wiped with isopropyl alcohol prior to application of the electrodes over the muscle. The subject will then lie down on a bed/platform which is suspended from the ceiling (approximately 10 inches from the floor), with the right leg extending over the end of the bed. The bed will be pulled back slightly, then released, causing it to swing towards the foot. At the lowest point in the swing, the foot will hit a vertical barrier, which will measure the amount of force impacting the foot. At the same time, a tiny accelerometer, strapped to the upper shin, will measure the acceleration being transmitted through the leg. The subject is then asked to contract the muscles of the anterior shin, which bring the foot and toes up, against an elastic resistance, until they are extremely tired. The impact will then be repeated.

On day 2, the same procedure will be repeated, but using the calf muscles, which point the toes down. A maximal contraction will be elicited with EMG analysis, followed by an initial impact. Fatiguing exercises will be done while lying on the bed, then a pendulum style impact will be repeated, while measuring impact force and acceleration.

POTENTIAL RISKS AND DISCOMFORTS

Although the impact force chosen will simulate that found in normal running, the knee and hip will be held straight, thus more force will be transmitted through to the trunk and perhaps the neck and head. Subjects with previous pain or swelling in the dominant leg, back, or neck should make it known to the research team.

Exercising the leg muscles to the point of maximal fatigue will result in mild discomfort at the time of the study. Muscle soreness and stiffness will probably be noticed the day after each testing session. This may last as long as 4 – 6 days.

The adhesive electrodes used for EMG analysis may cause a slight skin irritation. This will not last longer than a day or two and poses no significant health risk to the participant.

POTENTIAL BENEFITS TO SUBJECTS AND/OR SOCIETY

Subjects will benefit by gaining some insight into Kinesiology research.

Society will benefit by gaining a better understanding of the amount of shock force the body must absorb as the muscles fatigue. Also, how these absorption properties may differ between women of 20 – 25 years compared to women between 50 and 60 years of age.

PAYMENT FOR PARTICIPATION

Subjects will not be remunerated for participation in this study. They will be recruited on a volunteer basis.

CONFIDENTIALITY

Any information that is obtained in connection with this study and that can be identified with you will remain confidential and will be disclosed only with your permission. Only the researchers mentioned above will know your identity and personal information. This information will be stored in a secure computer in the ergonomics/biomechanics laboratory and will not be discussed or displayed in any form that would provide an indication of your identity.

PARTICIPATION AND WITHDRAWAL

You can choose whether or not to be in this study. If you volunteer to be in this study, you may withdraw at any time without consequences of any kind. You may also refuse to answer any questions you don't want to answer and still remain in the study. The investigator may withdraw you from this research if circumstances arise which warrant doing so.

RIGHTS OF RESEARCH SUBJECTS

You may withdraw your consent at any time and discontinue participation without penalty. This study has been reviewed and received ethics approval through the University of Windsor Research Ethics Board. If you have questions regarding your rights as a research subject, contact:

Research Ethics Co-ordinator
University of Windsor
Windsor, Ontario Medical
NpB 3P4

519-253-3000 X 3916

SIGNATURE OF RESEARCH SUBJECT/LEGAL REPRESENTATIVE

I understand the information provided for the study "*The effect of localized muscle fatigue on tibial shock acceleration, using the human pendulum method*" as described herein. My questions have been answered to my satisfaction, and I agree to participate in this study. I have been given a copy of this form.

Name of Subject (print)

Signature of Subject

Date

SIGNATURE OF INVESTIGATOR

In my judgment, the subject is voluntarily and knowingly giving informed consent to participate in this research study.

Signature of Investigator

Date

VITA AUCTORIS

Janice Flynn was born in Canada and graduated from the University of British Columbia with a Bachelor of Science in Rehabilitation in 1977. She has practiced physiotherapy in ten different cities, encompassing five Canadian provinces and England. Her return to academia was prompted by yet another move and an impending “empty nest”.